Development of Walk Assistive Orthoses for Elderly

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A l'élégance et à la démesure.

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Jeremy Olivier

Abstract

The proportion of elderly people is rapidly growing and the resources to help them will soon be insufficient. One of the main difficulties faced by the seniors is locomotion. Due to the heterogeneity of this population, several conditions may be responsible for gait impairment. Among them, the reduced muscular force is one of the most frequent. This thesis focuses on the design and the evaluation of new solutions for assisting people with reduced vigor. Robotic orthoses are then used to support critical movements required for walking.

Over the last two decades, the use of actuated orthotic devices for helping people suffering from gait disorders has been made possible. Recently, autonomous devices have even enabled spinal cord injured patients to walk again by mobilizing their paralyzed limbs. Addressing a completely different population, similar devices have been developed to augment healthy users' capabilities, for instance when heavy loads need to be carried. In this case, the wearer is in charge of the movements and the device simply follows the imposed trajectories. Extra load can then be carried by the exoskeleton without being perceived by the user.

The walk assistive devices developed as part of this thesis being intended for the elderly, they are at the intersection between these two classes of robotic orthoses. Indeed, most of the seniors who have difficulties to walk are able to move and therefore the mobilization devices are not adapted to them. Even though they need assistance, they surely do not want to have their movements imposed by a robotic device. The performance augmentation exoskeletons cannot help them either, as they simply follow the movements and only reject the external perturbations. A device that follows their movements (without constraining them) and that adds the right amount of force when needed is therefore required.

In order to achieve the demanding characteristics associated with assistive devices, new actuation solutions based on conventional electric motors are proposed. The combination of specifications in terms of overall weight, required assistance torque, dynamic capabilities or transparency when no support is provided is undeniably challenging. Various mechanisms are investigated to address these requirements. Two prototypes based on the proposed solutions are presented. The first one is based on a ball-screw transmission combined with linkages which provide a transmission ratio that is adapted to multiple walk related activities. The second one uses a transmission with clutches and an inversion mechanism which notably limits the losses due to the inertia of the actuation stage and greatly improves the natural transparency.

In order to limit the obstructiveness of the assistive device, we propose to use partial devices that support specific movements (e.g. flexion/extension of the hip). Two studies about the influence of such partial devices on gait are therefore presented. The first one focuses on identifying the potential sources of gait disturbance that orthotic devices can induce. The second examines the effects of an assistive controller implemented on one of the developed prototypes. These studies demonstrate that even though the passive influence of a hip assistive orthosis on kinematic patterns is limited, the metabolic cost is increased. A moderate assistance cannot compensate for this undesirable effect but a link between the hip assistance and the ankle trajectory could be established. This is of major importance as the elderly tend to rely more on their hip so as to compensate for their weak ankle muscles.

Key words: Actuated orthosis, Lower limb assistance, Influence of orthotic devices on gait, Actuation and transmission.

Résumé

La proportion de personnes âgées dans notre société ne cesse de grandir et les ressources pour encadrer ces gens viendront bientôt à manquer. Une des principales difficultés liée à l'âge est la locomotion. En effet, l'hétérogénéité de cette population implique que de multiples facteurs peuvent potentiellement détériorer la marche. Parmi ces derniers, la diminution de la tonicité musculaire est très fréquente. Cette thèse porte donc sur le développement et l'évaluation de nouvelles solutions orthétiques pour l'assistance de personnes souffrant de faiblesse musculaire.

Lors des deux dernières décennies, l'utilisation d'appareils orthétiques pour aider des personnes à mobilité réduite a vu le jour. De récents développements ont même permis à des patients paralysés de remarcher grâce à une mobilisation de leurs membres inférieurs. Dans un tout autre domaine, des appareils similaires ont montré leur aptitude à aider des personnes saines à augmenter leurs capacités nominales, par exemple lorsque de lourdes charges doivent être portées. Dans ce dernier cas, l'utilisateur bouge normalement et l'appareil suit simplement ses mouvements. Des charges peuvent ainsi être portées par l'exosquelette sans que l'utilisateur ne le perçoive.

Les appareils d'assistance à la marche qui ont été développés lors de cette thèse sont faits pour assister des personnes âgées et, en tant que tels, ils sont à l'intersection des deux catégories d'orthèses robotiques mentionnées plus haut. En effet, dans la majorité des cas, les gens qui ont des problèmes de locomotion liés à leur âge ne sont pas pour autant incapables de bouger leurs membres. Même s'ils ont besoin d'assistance, il est évident qu'ils ne veulent pas avoir un appareil robotique qui leur impose ses mouvements. Les exosquelettes permettant d'augmenter les capacités à déplacer de lourdes charges ne leur seraient guère plus utiles, étant donné qu'ils suivent simplement les mouvements et se contentent de rejeter les perturbations extérieures. L'orthèse d'assistance à la marche qui est requise pour cette population est donc un appareil qui suit les mouvements volontaires sans les contraindre et qui ajoute la quantité appropriée d'assistance lorsque cela est nécessaire.

Le cahier des charges en termes d'actionnement représente un défi en soi et de nouvelles solutions de transmission pour moteurs électriques sont donc proposées. En effet, la combinaison du poids réduit avec des couples d'assistance importants, des capacités dynamiques élevées et une transparence mécanique raisonnable lorsqu'aucune assistance n'est fournie est indéniablement difficile à atteindre. Plusieurs concepts sont donc présentés de manière à satisfaire ces multiples aspects simultanément. Deux prototypes basés sur ces concepts sont présentés. Le premier est basé sur une transmission à vis à billes fonctionnant avec un

mécanisme à barres. Cette dernière permet d'avoir un rapport de transmission variant avec la position et ainsi adresse de manière optimisée les besoins de plusieurs activités liées à la marche. Le second utilise une transmission munie de deux embrayages et d'un mécanisme d'inversion ce qui lui permet de limiter les pertes liées à l'inertie de l'actionneur et améliore significativement la transparence.
Dans le but de limiter la nature invasive que peut avoir une orthèse d'assistance, nous proposons d'utiliser des appareils partiels qui n'assistent que certains mouvements bien définis (p. ex. flexion/extension de la hanche). Deux études portant sur l'influence que peuvent avoir de tels appareils sur la marche sont présentées. La première vise à identifier les potentielles sources de perturbation que pourrait induire une orthèse d'assistance. La seconde se concentre sur les effets d'une assistance fournie au moyen d'un des prototypes développés. Ces études montrent que, même si l'influence d'une orthèse sur les trajectoires de marche reste limitée, le taux métabolique est quant à lui influencé négativement. Une assistance modérée ne permet pas de modifier ce constat ; cependant un lien clair entre l'assistance au niveau de la hanche et les trajectoires de la cheville a pu être établi. Ce dernier point est très important dans la mesure où les personnes âgées ont tendance à compenser leurs faiblesses du bas de la jambe par une activité accrue au niveau de leurs hanches.

Mots clefs : Orthèse actionnée, Assistance des membres inférieurs, Influence d'appareils orthétiques sur la marche, Actionnement et transmission.

Contents

List of Figures

List of Tables

1 General introduction

1.1 Motivation

With the population aging, the need for new technology to assist the seniors is growing. Indeed, their autonomy is usually diminished and the quality of their lives may therefore decrease (Kaye et al., 2000). Having a decent mobility is crucial both from the physical point of view as well as for psychological aspects. It has been reported that the aptitude to walk is one of the key points to be able to stay at home independently and to keep a decent physical condition (Paterson et al., 2004). In case of gait disorder, if a cane or a walker is not sufficient to assist a person in their everyday life, the unique remaining solution to keep a safe mobility is the wheelchair. This option represents additional and non-negligible psychological consequences (Nihei et al., 2008). Indeed elderly wheelchair users are more likely to have difficulties to perform activities of daily living and report as a majority that their health is fair or poor (Kaye et al., 2000). Moreover, physical condition may decline even more since using a wheelchair implies that the user will be in a sitting position most of the time.

The fall is the major threat for the elderly when they are walking. About 50% of the falls happen during some sort of locomotion (Winter, 1995). It is estimated that one third of the 65 years old and more tend to fall at least once a year (Moylan and Binder, 2007). This number grows even to one half for the 85 years old and more. The Swiss Federal Statistical Office (OFS) reported that about 25% of the 65 years old and more had fallen at least once during the last 15 months in 2007 (OFS, 2008). The consequences of falls are various (Moylan and Binder, 2007)(Lord et al., 2007). Fractures on legs (notably femoral neck), arms (radius or ulna), neck and trunk are common when falling. Head injuries, pain or superficial cuts and abrasions are also frequent. Falls also recurrently lead to depression and social isolation. Moreover, it is estimated that falls are responsible for proportionally nine times more deaths in the 80+ population than car accidents for the accident-prone 15–29 years old group (Winter, 1995). Falling being a central problem in elderly mobility, new solutions are required to reduce this risk.

Evidently, not all the causes of gait disorders affecting the elderly can be covered in this work. However, new practical solutions to promote mobility are proposed. This kind of developments is crucial as the lack of activity often initiates a vicious circle in which the elderly tend to walk less and consequently is one of the main causes for falling.

1.2 Structure of this thesis

This thesis focuses on the development of devices for helping the elderly to walk. Therefore the present chapter introduces important concepts about locomotion in general and about the elderly population. The problematic of walking for this specific population is stressed and various related issues are developed.
In chapter 2, a relatively broad overview on the orthotic devices for gait support is established. This chapter goes voluntarily beyond the main focus of this thesis as the approaches presented by some research groups certainly applies to the elderly population even though they were not necessarily intended so. This chapter is divided into two main parts. The first one focuses on the lower limb orthoses themselves while the second presents their control strategies.
Chapter 3 gives guidelines about what can be done to assist the seniors during walking and related activities (i.e. standing up from a sitting position, climbing stairs,etc.). Specifications for actual assistive walk orthoses are as well established in that chapter.
In chapters 4 and 5 mechanical concepts and actual solutions for assisting lower limb articulations are presented. The presented designs are based on the requirements that are developed in chapter 3.
Two studies realized with the developed orthoses are presented in chapter 6. The first one exhibits the effects of a passive device on gait. This study is of major importance as the negative impact of an assistive devices is often neglected even though it may overshadow its potential benefits. The second study is intended to illustrated the effect of a control strategy on gait.
Eventually in the last chapter, the presented developments and their implications for gait assistance are discussed. Considerations about future directions are also developed.

1.3 Human locomotion considerations

Locomotion is one of the most crucial Activity of Daily Living (ADL) and bipedal walk is the natural way the human beings accomplish this task. Other approaches may be adopted to move such as running, hopping or crawling but walking remains the most natural locomotion strategy for healthy adults. In absence of impairment, an individual is able to maintain his balance on his two feet and to move on various terrains to go where desired.
Depending on each person’s specificities, gait differences may be observed. This is mainly due to the differences in the musculoskeletal structure. As every individual optimizes his gait according to his own body (Anderson and Pandy, 2001), the resulting gait may differ slightly from one person to another. Other parameters influence gait in a more serious way. Some pathologies have a major effect on gait and this is further discussed in 1.4.2.
In this subsection some important definitions are reminded and the nominal human gait is described. The muscular activity acting on the main joints used during walking is also presented. A brief overview on the locomotive control is then introduced. Eventually, the

assessment techniques commonly used for evaluating gait are presented.

1.3.1 Gait parameters, phases and events terminology

A relatively important number of parameters are conventionally used to describe the human gait and the specific events that may be observed during it. The most important ones are explained hereafter.

Step and stride

The stride is defined by a complete walking cycle, which represents two steps. The position at the end of the cycle is then identical to the position at the beginning. Conventionally, the stride starts at the Right Heel Strike (RHS) or at the Right Toe Off (RTO) but it could start at any time during the cycle. Logically the stride length is the distance traveled during one cycle. It can be calculated for instance from the position of the right heel at the beginning and at the end of the stride. As the body posture is supposed to be identical at both moments, any body part can theoretically be used as reference. However, some body parts have a larger variability (e.g. the hands) and should therefore be avoided. The typical stride length of healthy male adults is $1.6\ m \pm 0.18\ m$ (Schmitt, 2007). The step length is measured from one step to the next. Note that it is not necessarily identical on both sides (and equal to half of the stride length) especially in case of pathologies affecting one side more than the other.

Cadence

The cadence is defined by the number of steps per minute and is directly linked to the cycle duration. With the cadence (or the cycle duration) and the stride length, the velocity can be calculated easily. The cadence is usually around $105\ steps/min \pm 7.7\ steps/min$ (Winter, 1984).

Stance, swing, single and double support

When walking, the two feet alternatively lift and move forward so as to enable a global displacement of the body. The phase during which a foot is in contact with the ground is called the **stance**. It typically lasts about 60% of the cycle and can be divided into three sub-phases. The first one is the early stance and starts at the heel strike and ends when the other foot takes off. The second phase is called **single support** and lasts as long as the second foot does not touch the ground. The third sub-phase is the push phase and ends at the Toe Off (TO). The **swing** is the phase during which the foot does not touch the ground and moves forward. As it lasts less than half of the cycle time, there is a period during which both feet are in contact with the ground and therefore form a closed kinematic loop. This phase is called the **double support**.

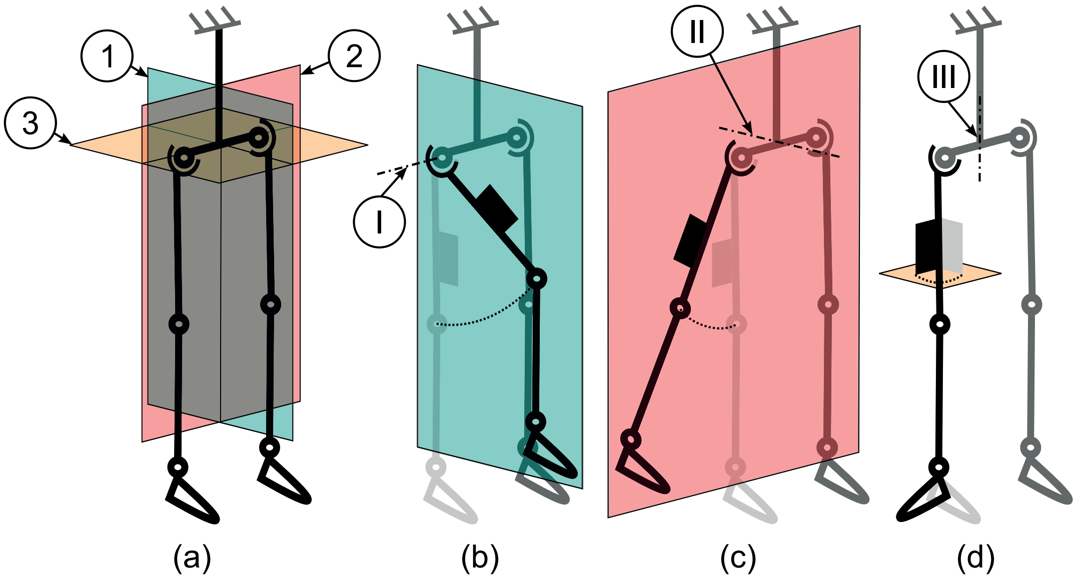


Figure 1.1 – Planes and axes used in biomechanics. (a) The first planes (turquoise) is the sagittal plane, the second (red) is the frontal plane and the third one (yellow) is the transverse plane. (b) The rotation about the transverse axis is the flexion/extension. (c) The adduction/abduction is about the antero-posterior axis and therefore acts in the frontal plane. (d) The internal/external rotation acts about the axis passing through the articulation and being parallel to the longitudinal axis.

Planes and reference axes

Three main planes are commonly used as references for the human body. They are orthogonal to each other and intersect at the body center of mass (see Fig. 1.1(a)).

The sagittal plane (1) is vertical and divides the body into two quasi-symmetrical parts (left and right).
The frontal plane (2) is also vertical but divides the body into the anterior (or ventral) and the posterior (or dorsal) parts.
The transverse plane (3) is horizontal (i.e. parallel with the ground). It divides the body into its lower and upper parts.
Usually theses planes are used to describe movements that act parallel to one of them. For instance, walking is often considered to be an activity taking place only in the sagittal plane. If the movement needs to be described more accurately, the projections of the various body parts on the other planes may also be used. However, as the planes are fixed, the description of a movement that combines a sequence of large angles (e.g. large hip flexion combined with abduction) may be misleading with such projections. A better way to describe precisely the movements could then be achieved with the techniques used by roboticists such as the Euler angles or the quaternions (Hamilton, 1850). The Euler angles have the advantage that they correspond to the commonly used articular angles when simple movements along one of the

reference planes are performed, which simplifies greatly their interpretation.
In addition to the planes, axes may be used to describe the body movements.
The transverse axis (I) is described by the intersection between the frontal and the transverse planes. In flight dynamics it corresponds to pitch .
The antero-posterior axis (II) is described by the intersection between the sagittal and the transverse planes. A rotation about this axis corresponds to roll in flight dynamics.
The longitudinal axis (III) is described by the intersection between the sagittal and the frontal planes. A rotation about this axis corresponds to yaw.

Articular angles

The terminology describing the angular movements of the different leg parts is usually common to all articulations even though exceptions exist.
Flexion and extension is the angle in the sagittal plane (see Fig. 1.1(b)). It may also be seen as a rotation about an axis parallel to the transverse axis. The flexion is the movement that reduces the angle between two body segments while the extension is the movement that increases it.
Adduction and abduction are the angular movements acting in the frontal plane (see Fig. 1.1(c)). The adduction is a rotation that brings a body part closer to the rest of the body and the abduction is logically in the other direction.
Internal and external rotations are the movements acting around their longitudinal axis (see Fig. 1.1(d)). An internal rotation brings the anterior part of a body part closer to the center line of the body while the external rotation moves it away.

Articulations of the legs

Three main articulations compose the leg: the hip, the knee and the ankle. A model of the lower limb kinematic chain is presented on Fig. 1.2. As the foot has many DoFs (notably due to the fact that the toes all have several mobilities) they are not shown on Fig. 1.2.
The **hip** is a spherical synovial articulation and therefore possesses three DoFs in rotation. Its movements may then be composed of flexion/extension, abduction/adduction and internal/external rotation. The Range of Motion (RoM) of the different movements are presented in Table 1.1 .
The **knee** is a 1 DoF articulation and enables only flexion/extension. Its RoM is presented in Table 1.1. However, it cannot be considered as a perfect hinge joint as, during the movements, the femoral condyles roll on the tibia and therefore the center of rotation moves. More complex models exist that describe the main movement of the knee e.g. (Sancisi and Parenti-Castelli, 2010). In addition, the neutral position of the knees (straight legs) may exhibit a permanent angle in the frontal plane. This angle is called varus in case the bowed legs and valgus in the case of the commonly called "knock-knees". Below the knee, the shank is able to rotate about

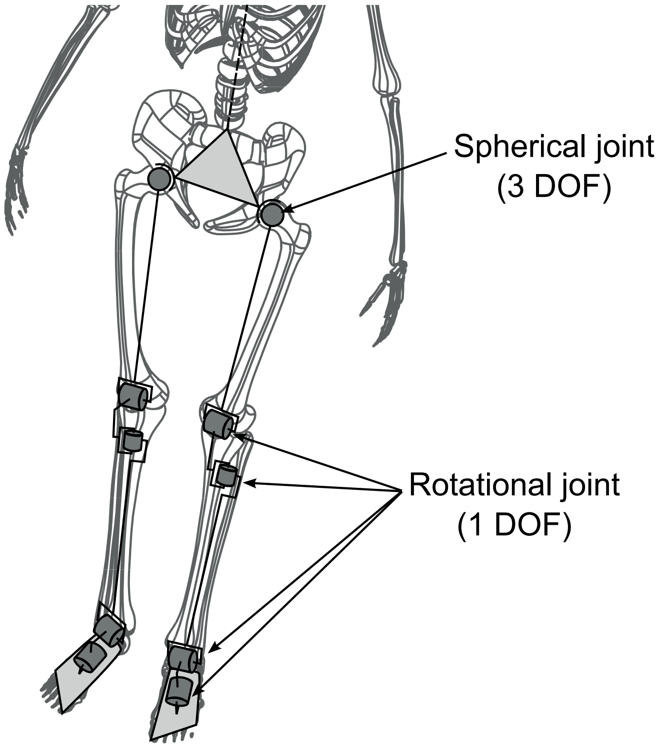


Figure 1.2 – Model of the leg kinematic chain. The hip is well approximated by a spherical joint. The knee is modeled as a rotational joint (as a first approximation). Another Degree of Freedom (DoF) is added below the knee to model the rotation of the shank in the transverse plane. The ankle is also a rotational joint and it enables flexion/extension. A second DoF corresponding to inversion/eversion is located close to the flexion joint. The latter is not associated only to the ankle as it also requires that some parts in the foot move.

Table 1.1 – Leg articulations RoM, adapted from (Roaas and Andersson, 1982). Note that according to Roaas, the hip possesses three rotations, the knee can only move in flexion/extension and the ankle has two DoFs. The rotation of the foot in the transverse plane is then not presented here.

Articulation	Motion	Max angle [°] mean	Max angle [°] std	RoM [°]
Hip	Extension	9.4	5.3	129.7
	Flexion	120.3	8.3	
	Abduction	38.8	7.0	69.3
	Adduction	30.5	7.3	
	Internal rotation	32.6	8.2	66.2
	External rotation	33.6	6.8	
Knee	Extension	−1.6	2.8	142.2
	Flexion	143.8	6.4	
Ankle	Dorsiflexion	15.3	5.8	55.0
	Plantar flexion	39.7	7.5	
	Eversion	27.6	4.6	55.3
	Inversion	27.7	6.9	

its longitudinal axis. With the upper part of the leg (thigh) fixed, the foot is then enabled to rotate on the transverse plane. This movement is called the tibial or foot internal/external rotation and is sometimes considered to be the second DoF of the knee.

The **ankle** also possesses one DoF: the dorsi/plantar flexion (which corresponds to flexion/extension for the other joints). This term is commonly used so as to avoid any confusion with the flexion/extension of the foot (toes). The other movement that is sometimes attributed to the ankle is the the inversion/eversion (which is a rotation in the frontal plane). This movement is in fact a combination of several articulations which involves the foot and the ankle. In general, the basic movements (i.e. acting parallel to one of the main planes) at the foot level involve combined (or polyarticular) actions as the axes of rotation are not all parallel to the three main axes defined above (antero-posterior, longitudinal and transverse).

1.3.2 Gait kinematics

Even though each individual has his own gait which can be considered unique (BenAbdelkader et al., 2004), a general tendency exists and standard trajectories can be identified for each joint. Many studies have been conducted to analyze those trajectories during walking (Johansson, 1973)(Winter, 1984)(Whittle, 1996). The pioneers in this domain worked with standard cameras and thus could only record the trajectories in one plane (usually the the sagittal plane). Recent techniques enable precise three-dimensional (3D) recording of each body segment (see 1.3.6).

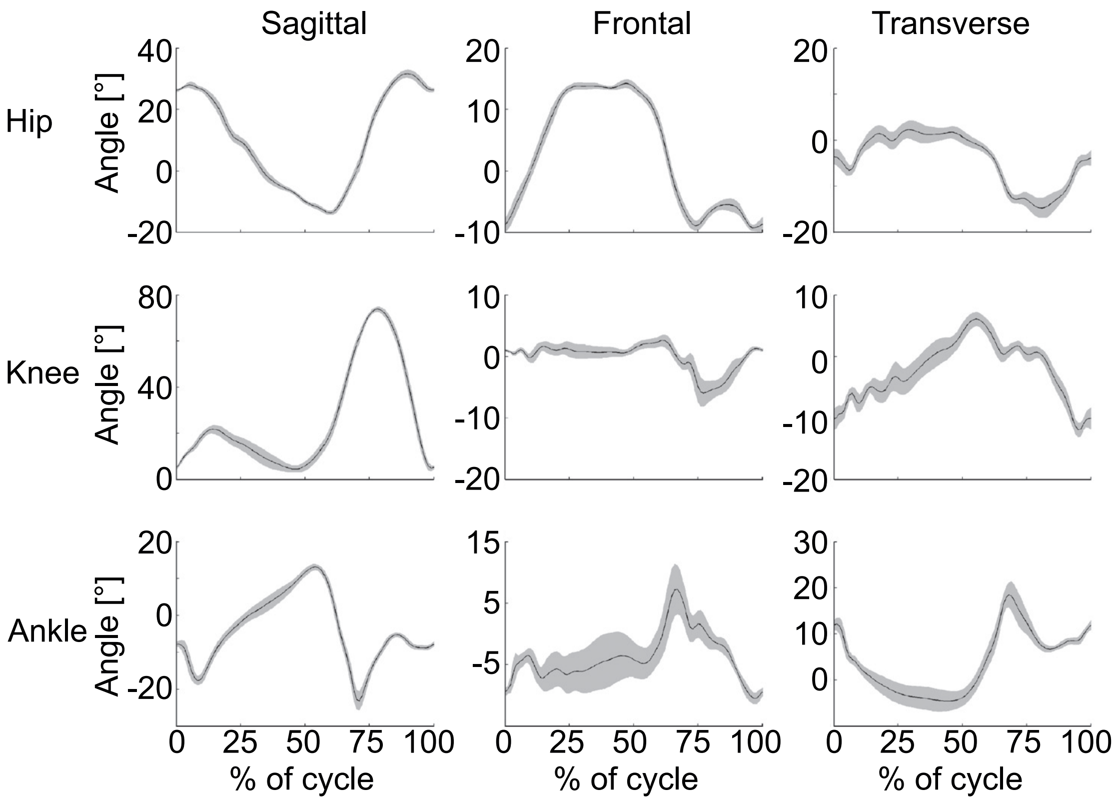


Figure 1.3 – Hip, knee and ankle joints trajectories during walking, adapted from (Mills et al., 2007). On this figure the cycle starts at Heel Strike (HS). The solid lines represent the mean profiles for one subject. The grey areas are the mean ±1 intra-subject standard deviation.

Among the numerous applications of such analysis two are of great interest as part of this thesis. The first benefit of such technique is that it enables to establish a diagnostic in case of pathologies and the second is that quantifiable assessment during rehabilitation is made possible. The curves shown on Fig. 1.3 illustrate the generic gait trajectories of a healthy subject joints. Evidently an inter-subject variability exists but typical patterns can easily be identified. Moreover, intra-variability is also present since human do not necessarily use a very stiff position control when walking and their movements are subject to perturbations (see subsection 1.3.5).

1.3.3 Gait kinetics

In order to follow the trajectories presented in 1.3.2, the articulations need to provide torques and forces. The torques acting on the different reference planes are presented on Fig. 1.4. **Inverse dynamics** is usually used to calculate those forces (Bresler and Frankel, 1950)(Eng and Winter, 1995). Assuming that the body segments are rigid, a multi-bodies system can be defined (Hardt and Mann, 1980). An example of such a system is presented on Fig. 1.5.

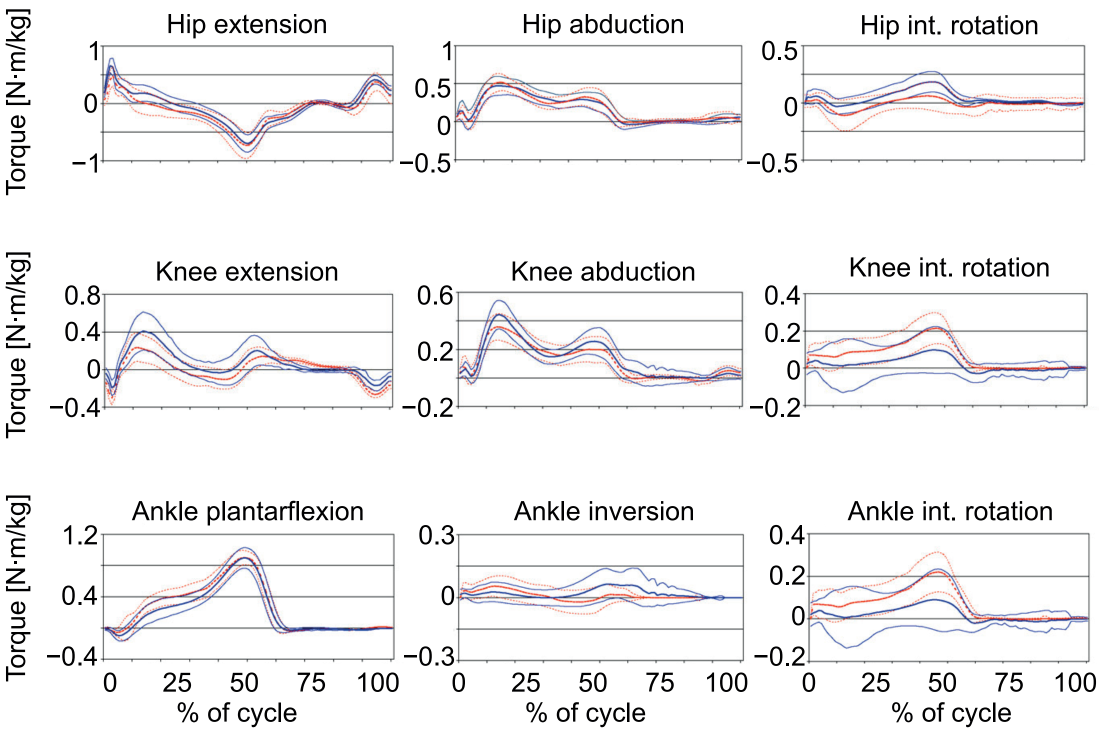


Figure 1.4 – Hip, knee and ankle joints torques during walking, adapted from (Riley et al., 2007). The torques are normalized by the weight of the subject. On this figure the cycle starts at HS. The blue thick solid lines represent the mean torques during overground walking and the lighter blue lines indicate ±1 standard deviation. The red dashed lines represent torques recorded on a treadmill.

In addition to the body parts trajectories, other parameters are required such as the Ground Reaction Forces (GRF) and the mass, the Center of Mass (CoM) and the inertia of each segment. The following equations are then used:

$$\sum \overrightarrow{F_{segment\_i}} = m_i \cdot \dot{\overrightarrow{v_{CoM}}} = \overrightarrow{F_i} + \overrightarrow{F_{i+1}} + m_i \cdot \overrightarrow{g} \tag{1.1}$$

$$\sum \overrightarrow{M_{CoM\_i}} = \frac{d(I_{CoM\_i} \cdot \overrightarrow{\omega_i})}{dt} = \overrightarrow{M_i} + \overrightarrow{M_{i+i}} + \overrightarrow{r_i} \times \overrightarrow{F_i} + \overrightarrow{r_{i+1}} \times \overrightarrow{F_{i+1}} \tag{1.2}$$

where $\overrightarrow{F_i}$, $\overrightarrow{F_{i+1}}$, $\overrightarrow{M_i}$, and $\overrightarrow{M_{i+1}}$ are respectively the forces and torques (3D vectors) applied at the two articulations (i and $i+1$) of the considered segment. $\overrightarrow{v_{CoM}}$ and $\overrightarrow{\omega_i}$ are respectively the translational and rotational velocities of the segment. m_i is the mass of the segment and I_{CoM_i} is its inertia. Note that the inertia must be considered according to the rotational velocity. $\overrightarrow{r_i}$ and $\overrightarrow{r_{i+1}}$ are the position vectors between the CoM and the two articulations. "×" indicates a cross product. The same equations apply for each segment having two articulations. In case of the last link in the kinematic chain and if it is not touching anything, one of the couples $(\overrightarrow{F_j}; \overrightarrow{M_j})$ is a null vector (six-dimensional vector). This information can be used instead of the force plate during the single support phases. If a segment has more than two articulations, additional couples $(\overrightarrow{F_k}; \overrightarrow{M_k})$ can be added.

1.3.4 Muscular activity

The voluntary torques applied at the articulations are generated by the muscles. Even though the complexity of the muscles system goes beyond the the scope of this thesis, some basic aspects of the lower limb system is described here. Unlike conventional robotic systems whose DoFs are actuated each by a single motor, the biological joints require the action of several muscles. The first reason for this is that muscles can only produce force when they contract. Antagonist actuation units are then necessary. At least two muscles are required to perform a movement in both directions (e.g. flexion and extension). As the muscles and tendons have a non-linear springs behavior, co-contraction can be used to modulate the stiffness of a joint (for more details about this mechanism, see 4.1.3). This stiffness variation may be used to take advantage of passive dynamics effect or to adapt the limb behavior to unexpected events (Voloshina et al., 2013). For instance, during walking the knee usually has a compliant behavior during the swing phase while a higher stiffness is required during stance (Pfeifer et al., 2014). The stiffness at the ankle level plays also a great role when walking on uneven terrains (Voloshina et al., 2013). In addition, some muscles have an effect on more than one joint. For example, the quadriceps have an effect on the knee but also on the hip. The main muscles responsible for the movements are presented on Table 1.2. A simplified version with only seven muscle groups (Vaughan et al., 1992) can be used for describing the walking cycle (see Fig. 1.6). Fig. 1.7 shows the sequence of activation of these muscle groups.

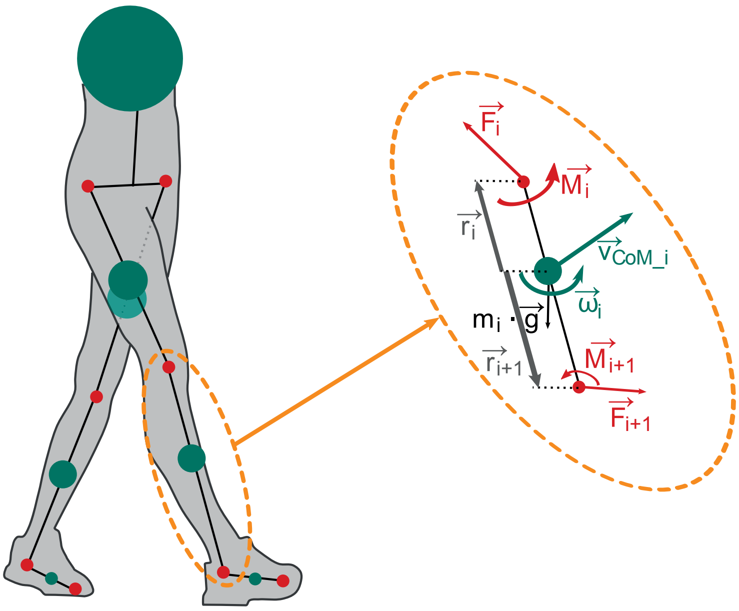


Figure 1.5 – Multi-bodies system representation of the human body. In this approximation, each segment has two articulations (red dots). The CoM of each segment is shown in turquoise (the size of the dot is proportional to the segment mass). The parameters used in equations (1.1) and (1.2) are detailed for one segment (on the right). Note that the torques and rotational velocities are represented with curved arrows and not with actual vectors in order to differentiate them from forces or displacements.

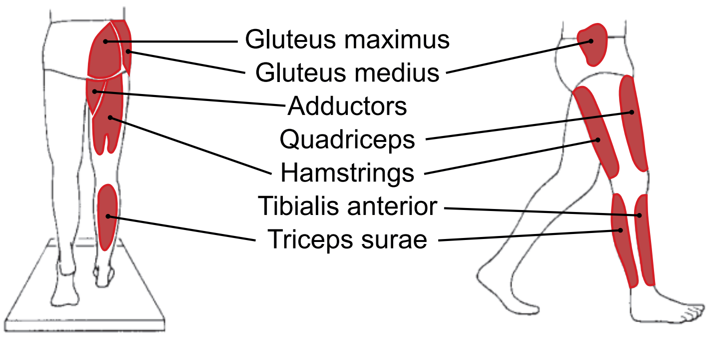


Figure 1.6 – Seven major muscle groups of the lower limb, adapted from (Vaughan et al., 1992).

Table 1.2 – Main muscles required to perform specific movements, adapted from *http://www.exrx.net/*. The muscles in bold are shown on Fig. 1.6.

Articulation	Motion	Main Muscles
Hip	Extension	**Gluteus maximus** **Semitendinosus*** Semimembranosus Biceps femoris (long head)
	Flexion	Iliopsoas Tensor fasciae latae **Rectus femoris****
	Abduction	**Gluteus medius** Gluteus minimus
	Adduction	**Adductors**
	Internal rotation	Tensor fasciae latae **Gluteus medius** Gluteus minimus, anterior fibers
	External rotation	Gemellus superior Gemellus inferior Obturator internus Obturator externus Quadratus femoris
Knee	Extension	**Quadriceps femoris**
	Flexion	**Hamstrings**
	Internal rotation	Popliteus Semimembranosus Semitendinosus
	External rotation	Biceps femoris
Ankle	Dorsiflexion	**Tibialis anterior**
	Plantar flexion	**Gastrocnemius***** **Soleus*****

*part of the hamstrings **part of the quadriceps ***part of the triceps surae

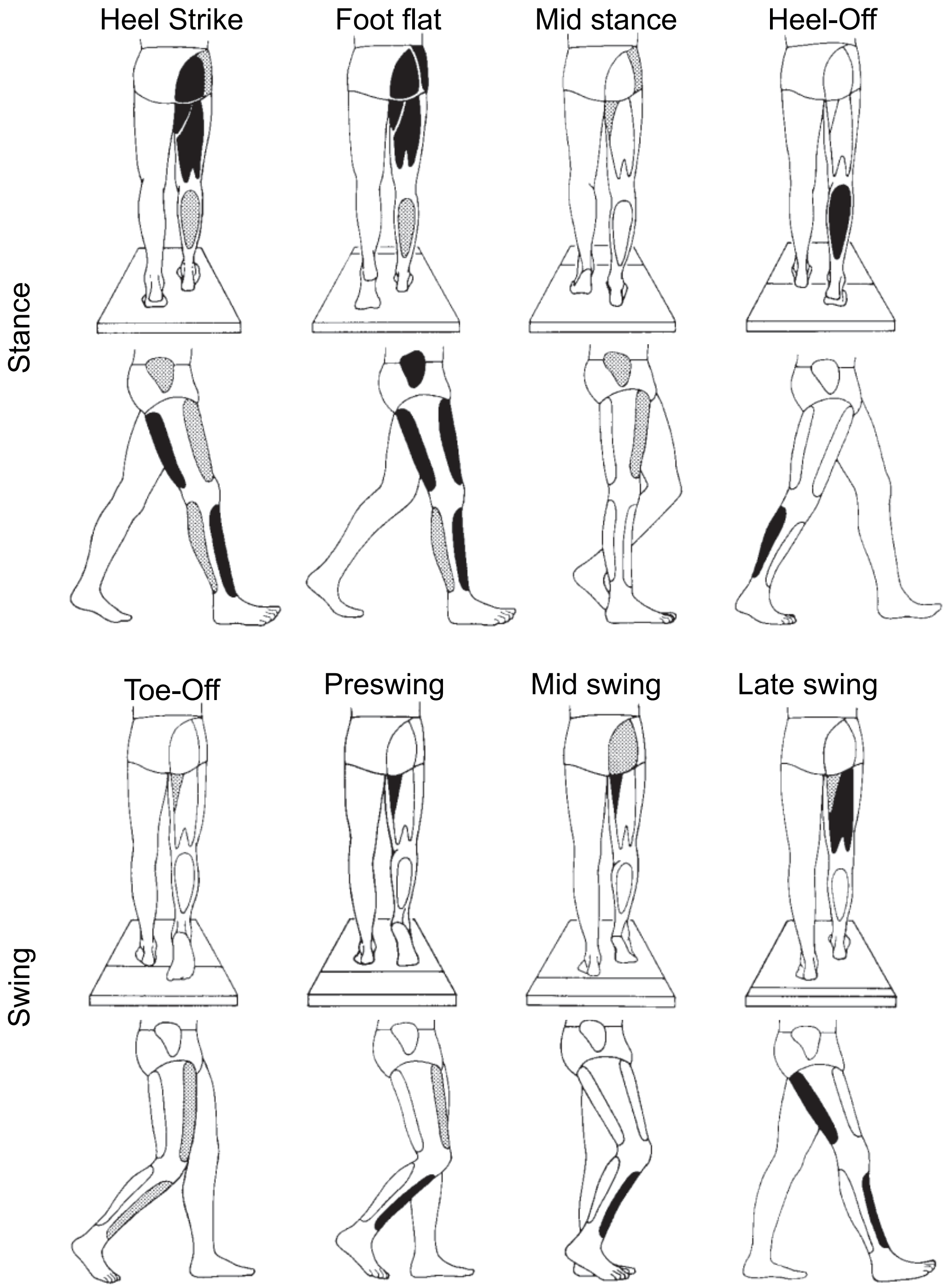


Figure 1.7 – Activation of the major muscle groups of the lower extremity during walking, adapted from (Vaughan et al., 1992). The shading of the muscle groups represent their activation (black being the most active).

1.3.5 Locomotion control

The coordination of the muscle activation is managed by the Central Nervous System (CNS) (Duysens and Crommert, 1998). Efferent nerves drive the motor signals to the muscles while afferent nerves provide feedback from the various body sensors (muscle spindles, Golgi tendons, mechanoreceptors, etc.). The system works then in closed-loop. When a task is to be performed, the CNS organizes which joints have to be moved, where and when. In the case of locomotion, neural networks are in charge of generating patterns which enable the body to perform complex cyclic tasks involving multiple joints. A widely accepted theory states that the so-called Central Pattern Generator (CPG) initiates the basic motor trajectories at the spine level (Loeb, 1989)(Crommert et al., 1998)(Zehr, 2005). On top of that, the initiation (and the termination) of the rhythmic nervous signals are initiated at the supra-spinal levels (pre-motor and motor cortex and cerebellum). Another source of modulation comes from reflexes (Nielsen and Sinkjær, 2002). Under normal conditions, the reflexive pathways increase the efficiency of gait and when unexpected perturbations arise, reflexes help to stabilize the posture. The information coming from the afferent nerves is also sent to the brain where it is combined with other sensory inputs (visual, auditory, vestibular,...). The high-level control may therefore also benefit from those inputs. Fig. 1.8 illustrates how the sensroy-motor system works in humans.

1.3.6 Assessment techniques

In order to assess the quality of locomotion, several methods may be used. Some of these techniques are based on tests that follow precise protocols. The goal of these tests is usually to evaluate if/how a person is able to perform a particular activity related to locomotion.
Based on what is presented above, objective characteristics can also be identified and measured. Kinematic and kinetic parameters are the most common data that are collected when assessing gait quality.
A third category of test exists that measures the subject's physiological signals such as the muscular activity or the energy expenditure. This is a valuable information as it describes what is happening at the subject's level.

Performance based assessment

The performance based tests assess the locomotion by investigating the subject's ability when performing specific tasks. Usually these tests are performed on elderly subjects or patients suffering from Parkinson, multiple sclerosis, hip fracture, Alzheimer's, stroke, etc. The traditional neuro-muscular examination appears to be a poor indicator of functional capabilities (Podsiadlo and Richardson, 1991). Therefore, the performance based tests are not based on the detection of specific symptoms or impairments (e.g. tension problems, diminished muscular force, poor eyesight, etc.) but rather on the ability to complete ADL. One of the main advantages of such tests is their simplicity. Indeed, they can be done in a relatively

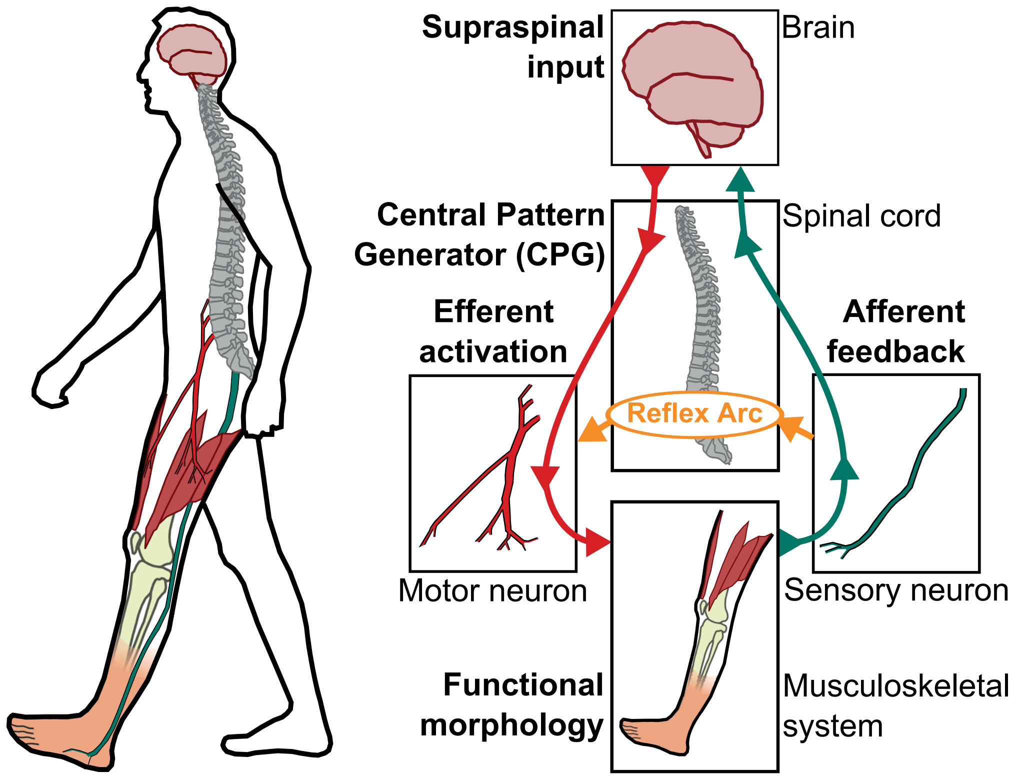


Figure 1.8 – Sensory-motor control loop, adapted from (Tucker et al., 2015). The CPG initiates the locomotion cyclic trajectories, which implies synergistic actions of multiple muscles. The motion intentions are initiated at the supra-spinal level. A volitional modulation may also be generated at any time at this level. The muscles are activated through the efferent nerves. The afferent pathways are used to send the information from the various receptors to the spine and the brain. Reflexes can then be used to rapidly react to unexpected event.

short time and no complex equipment is required.
One of the most used tests is the Tinetti Performance Oriented Mobility Assessment (POMA). It is composed of two distinct parts. The first one is a balance assessment where the static balance (sitting and standing) and the sit-to-stand and stand-to-sit transitions are analyzed. The second part focuses on gait. Specific characteristics such as the step length and height or their symmetry are evaluated. For a complete description, see (Tinetti and Haven, 1986).
Another very common test is the Timed Up and Go (TUG). This test assesses the ability of a person to stand up from an armchair, walk three meters, turn around, walk back and sit down. This needs to be done when asked by the examiner and without using any help. The only used metrics is the time that is required. A score of 30 s or more indicates that the subject is prone to falls (Podsiadlo and Richardson, 1991).

Kinematics and kinetics assessment

The articulations trajectories are a valuable information to detect gait disorders. For instance, compensating movements or limitation in RoM are easily detected with such motion tracking methods. Moreover, precise measurements may help to study the evolution of a disease affecting gait. A large variety of human **motion capture** systems exists (Zhou and Hu, 2008). The most used techniques are concisely presented hereafter.
Three different strategies may be adopted to track the movements of the body parts. First, vision based methods can be used. The second category of motion tracking relies on wearable sensors. The third one consists in using a structure attached to the body. The positions of the structure parts are then used to deduce the posture of the human. Note that these systems have a lot in common with the exoskeletons that do not constrain the movements (see 4.2).
Historically the first gait recordings were performed with standard cameras. With the latest progress in image processing, complex **optical motion tracking** systems using more than one camera have arisen. With such systems a 3D reconstruction is made possible (using triangulation). Passive (e.g. Vicon, *UK*, OptiTrack, *US* or Qualisys, *Sweden*) or active markers (see Appendix A) are usually required but systems without markers also exist (e.g. Microsoft Kinect, *USA*). The latter use algorithms designed to recognize specific shapes (e.g. human body parts).
The main problem of the vision based techniques is that they are subject to occlusions. One possible alternative is proposed by **magnetic motion tracking** systems. Other problem arise with magnetic systems such as the distortion of the measured field or the latency (Zhou and Hu, 2008). Moreover, the area in which the measurement is possible is bounded (similarly to the optical tracking systems).
Methods based on **Inertial Measurement Units (IMUs)** have therefore became popular for tracking the movements in daily life environments (Tao et al., 2012)(Aminian and Najafi, 2004).
The torques and forces applied at the joint level are also important for gait analysis. As explained in 1.3.3, if precise measurements are desired, GRF are required (at least during the double support phases) in addition to the trajectories, the mass and inertia of the different body segments. These GRF are usually measured with **force plates**. Conventionally the force

plate is stationary and the subject needs to step on it, which implies either that only one stance phase can be measured or that multiple force plates are required. An alternative is however possible with force sensors integrated in the subject's soles (Liu et al., 2010).

Physiological signals assessment

As explained in 1.3.4 the torques at the joints level are the result of multiple muscles actions. The stiffness can, for instance, be modulated by the co-contraction of antagonist muscles. Moreover, the **action of the muscles** is constantly optimized by the CNS according to the performed activity (Franklin et al., 2008). Consequently, knowledge about the articulation torques only does not guarantee a complete understanding of the underlying gait mechanisms. A deeper analysis at the muscles level may then be required. Electromyography (EMG) is the most commonly used method for measuring the muscular activity even though some other techniques exist (e.g. Mecanomyography (MMG), or measurement of muscle volume (Kong and Jeon, 2006) or hardness (Lukowicz et al., 2006)). This technique is usually very efficient for detecting the activation onsets or the rest periods, which is crucial for analyzing the coordination. However the amplitude of the activation is more challenging to interpret as the quantitative relation between EMG and muscle tension is less clear when the muscles change in length (i.e. eccentric and concentric activations) (Inman et al., 1952)(Kuriki et al., 2012).
Another commonly used metric is the **metabolic energy cost**. This measurement gives general information about the adaptation of an individual to various situations (Wezenberg et al., 2011) or to a specific pathology (Waters and Mulroy, 1999). This is done by looking at the oxygen consumed during an activity (milliliters of O_2 per kilogram body weight per minute). Even though the repeatability from one day to another is not extremely reliable, the comparison between activities performed in a short period is clearly valid. In that case the effect of fatigue needs to be estimated carefully (for instance by comparing similar conditions at the beginning and at the end of the assessment). In absence of cardiac disease, a close relationship between the oxygen uptake and the heart rate exists. It can then also be used as an indirect measurement of the metabolic energy consumption.

1.4 Walking impairments in elderly

In theory and in absence of pathology, the seniors have gait patterns that are very comparable to the younger people's when walking at the same velocity (even though few exceptions exist). Indeed, Jansen et al. highlighted that if only healthy and active seniors are considered, there is no significant difference with younger subjects in terms of main gait parameters (i.e. cadence, stride length, double-support time, etc.) (Jansen et al., 1982). However a vast majority of the seniors have a longer reaction time, reduced visual and audio capacity and they tend to have a higher energy consumption when walking. They also have an increased sway while standing (Sheldon, 1963), and they require more attentional resources to recover from an external perturbation (Brown et al., 1999). All these factors have an evident impact on how the elderly apprehend walking as well as on the risk of fall. Moreover, due to their longer medical history, most seniors tends to exhibit notable differences in terms of gait parameters. A norm is then difficult to establish as the elderly population is extremely heterogeneous. This implies that many different causes for gait disorders exist.

1.4.1 Gait alteration in absence of directly related pathology

Conservative gait pattern

Even in absence of pathology in direct link with gait, walking pattern adaptations tend to appear with age. Indeed, the elderly are very likely to reduce their natural pace. The normal velocity of a fit person at the age of 80 is usually 10 to 20% slower than that of a younger person (Desforges and Sudarsky, 1990). Along with this they usually have shorter steps, a reduce cadence and a longer double-support phase (Murray et al., 1969). A common interpretation of these changes is that the elderly tend to adopt a more conservative gait pattern (Menz et al., 2003). This strategy is supposed to limit the accelerations at the trunk and head levels. This has then a positive impact on the vision and the vestibular system which are crucial for balance, especially in elderly.

Gait changes associated to aging

The gait adaptations in direct link with the reduced velocity are multiple. Logically most of them disappear when walking at a higher pace as stated by (Jansen et al., 1982). However, some of the adjustments persist as highlighted notably in (Kerrigan et al., 1998). Reduced hip extension, changes in the lower trunk coordination and reduced ankle plantar flexion and ankle power generation are the characteristics that turn out to be related with age and not only due to the reduced walking velocity.
The reduction of the hip extension is attributed to a significant **hip tightness**. Kerrigan observed that it is even more pronounced in elderly with a history of falls (Kerrigan et al., 2001). The hip tightness may also contribute to the change in anterior pelvis tilt. Indeed this postural change is believed to be a compensation strategy that is induced by the limited hip extension capability (Kerrigan et al., 1998). McGibbon explained the changes in the lower trunk by highlighting the difference in trunk and pelvis phase shift (McGibbon and Krebs, 2001). Indeed the younger subjects tend to lead the movement with their pelvis while the elderly prefer a trunk-leading strategy. As a consequence, the mechanical energy expenditure that they need to provide with their low-back musculature is increased. McGibbon suggests that this may be partially due to a strategy where concentric muscle activity is diminished as it is believed to be more metabolically demanding than eccentric activation[1].

[1]A concentric muscle activity means that energy is generated. Conversely, during eccentric muscle activity energy is dissipated.

Another factor that could explain partially this alteration is the **diminished musculature at lower extremities** especially at the ankle. The pelvis and trunk muscles are then used to pull the leg into swing as the ankle plantar flexors (and to a lesser extent, the knee extensors) are too weak to perform a vigorous push-off. This gait alteration is often referred to as the **hip-strategy** as opposed to the ankle-strategy. This explanation goes along with the observation that the elderly have difficulties to generate high ankle power (Judge et al., 1996). This was revealed by recording the joints power when elderly people walk at their maximal pace. In these conditions they are usually unable to increase their ankle plantar flexion power as would younger people do. The direct consequence is that the plantar flexion is diminished but in turn more power needs to be generated at the hip level (flexion).
Psychological effects also exist. The fear of falling for instance has been pointed out as a source of gait changes (Reelick et al., 2009). Moreover people with this kind of fear tend to limit their activity and therefore become weaker which has an dramatic impact on their autonomy. The other psychological/psychiatric conditions that have a strong correlation with gait and balance disorders are: depression, sleep disorders and substance abuse (Salzman, 2010).

1.4.2 Pathologies affecting gait

In addition to the common gait adaptations due to age, a large number of pathologies affecting mostly the elderly may have a direct impact on the ability to walk. The cardiovascular diseases (e.g. arrhythmias, orthostatic hypotension, congestive heart failure, coronary artery disease,...) evidently have an impact on the effort that a person can provide and therefore on gait (Salzman, 2010). Moreover, as walking requires precise coordination of a large number of muscles, pathologies affecting the nervous and the musculoskeletal systems usually have the most serious consequences on locomotion (Salzman, 2010). However, precise data on the prevalence of the different pathologies and their relation with gait impairment are difficult to find. Some studies have been conducted in medical centers but they do not accurately represent the global population (Stolze et al., 2005)(Sudarski, 2001)(Verghese et al., 2006). Moreover, gait disorders are often caused by more than one factor (Sudarski, 2001). This subsection therefore concisely presents some of the most referenced pathologies affecting gait. It is divided in two categories: the neurological and the musculoskeletal diseases.

Neurological pathologies

Among the neurological pathologies affecting elderly, **stroke** is the most common cause of gait disorder. This pathology, also called cerebrovascular accident (CVA), is the result of a poor blood flow in the brain, which causes cell destruction (Donnan et al., 2008). Evidently the areas where cells are dead are not functioning properly and this may affect gait (typically in 60% of the cases, (Stolze et al., 2005)). Patient with post-stroke hemiparesis have a reduced progression velocity, a slower cadence, shorter stride length and asymmetries in spatial, kinematic and kinetic gait variables (Chen et al., 2005). The ankle push-off of the paretic leg is notably reduced. This induces compensation strategies which in turn augment the energetic cost of walking. The balance is also affected and the stroke patients tend to compensate with a greater step width.

Idiopathic Parkinson's Disease (IPD) affects less people but its effect on gait is proportionally more frequent than that of stroke. In (Stolze et al., 2005), 93% of the people suffering from IPD had gait impairment. IPD is a degenerative pathology affecting the CNS and therefore directly alters the movements. The most common effect of IPD (about 80% of the cases) is the "slowness in the performance of movement sequences" (Morris, 2000). The very severe case where the movements are dramatically restrained is called "hypokinesia". The less dramatic cases are refereed to as "bradykinesia". The people suffering from hypokinesia walk with short strides, limited arm swing (usually more pronounced on one side) and reduced trunk rotation. Other effects of IPD may arise such as the "akinesia" which is an inablility to initiate a movement or the "freezing" which is a sudden cessation of a movement during the execution of a sequence of actions. These effects appear to be context dependent and may typically happen when approaching a door or when stepping on a different surface on the ground (e.g a carpet). IPD may also provoke difficulty to terminate locomotion, which evidently may have severe consequences for example when approaching stairs.

Myelopathy is a deficit affecting the spinal cord and is common in elderly (Sudarski, 2001). Due to a mechanical compression (caused for instance by ligament hypertrophy or by arthritic changes of the cervical spine), motor and sensory capacity are altered. The gait of people suffering from myelopathy is characterized by "clumsy movements" (stiffness and bounce) in the legs, an inclination to circumduct[2], slower pace and longer double support phases (Sudarski, 2001). In (Malone et al., 2012), the kinematic analysis showed notable differences at the pelvis (reduced obliquity and rotation RoM), the hips (reduced flexion/extension and adduction/abduction RoM and different internal/external rotation pattern), the knees (reduced peak flexion both in stance and swing phases) and the ankles (reduced peak plantar flexion). The kinetic analysis also exhibits differences at the three articulations with globally reduced peak torques and modified patterns. Here again some of the variations are due to the reduced pace but the differences at the knee and ankle remain significant even at matched walking velocity. Interestingly, this population does not seem to be able to compensate for the lack of power at the ankle level with higher hip flexion torque.

People suffering from diabetes mellitus (DM) and **peripheral neuropathy (PN)** are also subject to gait disorders (Richardson et al., 2004). Diabetes is a group of metabolic diseases and not directly a neurological disorder. However it may have an indirect effect on the nervous system. Due to defects in insulin secretion or to insulin action, hyperglycemia appears (American-Diabetes-Association, 2013). PN is thus commonly caused by DM even though some other pathologies may be responsible for it (e.g. vitamin deficiency, excessive alcohol consumption, etc.). Cavanagh et al. reported that the incidence of fall in DM patients with PN is about 15 times higher than in a control group composed of diabetic patients without PN (Cavanagh et al., 1992). PN can then be considered as the actual cause of gait disorders in diabetic patients. Richardson et al. noted that the older women with PN reduce their step

[2] circumduction is combination of flexion/extension and adduction/abduction

length which certainly compromises the gait efficiency. Mueller and colleagues demonstrated that people suffering from PN rely much more on a hip-strategy than on the plantar-flexor muscle (ankle-strategy). Note that in this study the mean age was relatively low (mean age: 57.3 years old with a standard deviation of 12.6 years). As the elderly already have a tendency to rely more on the hip than on the ankle in absence pathology (see 1.4.1), the effect of PN is thought to be even more pronounced for this part of the population. The reason for this strategy modification in PN is explained by the the need to reduce the GRF; especially the shear forces that are particularly damageable for an insensitive foot (Mueller et al., 1994). Moreover, a lighter push-off implies that the GRF are spread over a larger surface as in that case not only the forefoot is in contact with the ground.

Musculoskeletal pathologies

Osteoarthritis (OA) is a degenerative disease which affects the cartilage and its surrounding tissues (Litwic et al., 2013). As the constraints due to weight-bearing on the knees and the hips are important, they are often considered to be the most critical articulations. The feet may also be subject to OA. The symptoms associated with this disease are joint pain, stiffness and limitation of movement (Litwic et al., 2013). According to Litwic and colleagues, nearly 50% of the people aged 75 years old and more suffer from OA. The statistics concerning the hip are less serious but remain high as 17% of this part of the population is affected. The consequences of OA on gait are multiple. The first observation is that the severity of the disease is well correlated with the walking velocity (Zeni and Higginson, 2009). Then, the stride length is usually reduced and the stance phase lasts longer (Al-Zahrani and Bakheit, 2002). The RoM of all the leg articulations are also affected. Zeni and Higginson reported that at self-selected speed, the torques in the sagittal plane at the knee and at the ankle level tend to decrease while the torques at the hip follow the inverse trend (Zeni and Higginson, 2009). The reduction of the progression velocity can easily explain the torque reductions at the knee and the ankle but certainly not the augmentation at the hip. The knee torque in the frontal plane also increases with the severity level of OA. Analysis on the muscle activation demonstrated that the coordination is also altered by OA (Childs et al., 2004). A significant increase of the co-contraction was notably observed. Childs et al. attribute this effect to an attempt to reduce pain and/or to stabilize the knee during the weight bearing phase.

Sarcopenia is another very common condition affecting 30% of individuals over 60 years of age and more than 50% of people over 80 years (Baumgartner et al., 1998). Sarcopenia is not a disease as such but is a consequence of multiple factors such as changes in the nervous and hormonal systems and sedentary lifestyle (Vandervoort, 2002). It is described as a low relative muscle mass and is associated with metabolic, physiologic and functional changes in elderly. Its classification in this subsection is therefore controversial as muscle weakness/atrophy could be attributed to normal aging considering the proportion of affected population. Many authors include sarcopenia in normal aging and for this reason its main effects on gait are already described in 1.4.1. Note that here again the multifactorial nature of the gait alteration in elderly makes it difficult to differentiate the individual contribution of each pathology. However the reduced capabilities of the muscles convincingly explains why a redistribution of the joint torques appears (DeVita and Hortobagyi, 2000).

1.4.3 Risk of falling

The factors leading to falls in elderly are multiple (Lord et al., 2007). In that context the ability to maintain a vertical position is a key element. The seniors postural stability is often decreased due to factors such as diminished muscular force, decreased vision, affected vestibular function, loss of sensitivity in peripheral sensory systems or difficulty to process rapidly and adequately the afferent inputs (Lord et al., 2007).
Although necessary to keep balance, avoid obstacles or detect and react appropriately to hazards, these components are not sufficient to ensure safety in ADL. Indeed, a majority of falls happen during walking and walk related situations (Berg et al., 1997). The articulation RoM, the tension problems, the lack of attention, the possible confusion, the medication or non-adapted environments have then also a major influence. Other situational factors play a role on the risk of falling. Carrying objects for instance is strongly correlated with it (Nachreiner et al., 2007). Another critical situation is the negotiation of stairs. Even though it does not represent the majority of falls, stair climbing/descending is responsible for a large proportion of the most critical ones (Lord et al., 2007). In this situation the foot placement is critical and an error may easily lead to loss of balance.
During walking two main types of mechanisms may lead to falls. The first one is tripping. It appears that the motor control as well as the muscular capacity play a major role in recovering from trips. Pijnappels and colleagues highlighted that elderly people who are not able to recover from tripping have a decreased capability to reduce rapidly the torque at the lower limb joints and a significantly lower peak ankle torque when tripping (Pijnappels et al., 2005). The second gait-related hazard is slipping. Usually slips happen shortly after HS. Here again an appropriate strategy is required so as to avoid loss of balance. Studies have shown that the elderly usually need more time to initiate a response to slip (Lord et al., 2007). Moreover, due to a lack of muscular vigor they are not able to produce as much torque as young people do. The duration of the muscular activity is also longer and the secondary strategies are more exaggerated. The latter are notably the trunk and arms movement responses.
Fall is also frequently associated with the sit-to-stand transitions (Pai et al., 1994). The strategies involved in this task imply mainly the ability to maintain the projection of the center of mass inside the base of support. If it is not the case at a certain point in time, the horizontal momentum of the center of mass needs to be directed towards the base support. The magnitude of the momentum needs to be sufficient so as to reach an eventual stable equilibrium. Enough strength and appropriate control are thus the two requisites conditions to fulfill the task. Logically, IPD and stroke patients are typically prone to have difficulties to achieve proper sit-to-stand transitions (Inkster et al., 2003)(Lomaglio and Eng, 2005).
In essence, to avoid a fall, three main conditions need to be fulfilled (Lord et al., 2007). The

first one is the detection of a potential hazard. Once the threat is identified, a correct response is to be selected. Eventually, the response needs to be executed properly. With the possible impairments at the sensory, the cognitive and motor levels, the elderly are logically more likely to miss one of the conditions and therefore to fall.

1.4.4 Discussion

Aging implies that a large number of factors can affect gait. During walking, the muscles control the articulations which move so as to perform the desired trajectories. The different sensors indicate to the CNS the various data that are useful for performing safe actions. According to the conditions a strategy is decided at the supra-spinal level. Reflex loops modulate the overall execution in case of specific events. Considering this, it is evident that many disorders can generate gait troubles. The articulations (cartilage, ligaments, bones), the muscles, the various nerves for sensory or motor tasks, the spine, or even the brain may then be affected by some sort of affliction.

When a gait disorder appears, an appropriate diagnosis is mandatory so as to localize the source(s) of the problem. According to this appreciation, an action corresponding to the level of the impairment can be implemented. Rehabilitation therapy is of course preferred in the first instance. However, not all troubles can be fully cured with such treatments and in those cases alternatives are required.

Robotic orthoses are potential candidates to address gait disorders as they can act at the three possibly impaired levels: sensory, motor and control. In the case of muscular issues, the problem can *a priori* be addressed relatively straightforwardly by supporting (or supplanting) the motor units. On the other hand, a trouble affecting high-level functions (i.e. brain or spine disorders) requires more evolved procedures. If a pathology affects the spine[3] for instance, the generation of trajectory needs to be assisted (this already requires a low level control loop) but the intention also needs to be decoded (mid or even high level). In this case the robotic device would need to assist the sensory, the motor and part of the control actions at a very high degree (for further details refer to 2.2).

The present work therefore proposes robotic orthoses adapted to the elderly. This population being extremely heterogeneous, the sub-category of people suffering from muscle weakness is targeted. Their needs differing from those of other populations for whom orthotic devices exist, adapted specifications evidently need to be established. From these, new solutions are elaborated.

[3]The most extreme case would be a complete lesion of the spinal cord interrupting both afferent and efferent pathways.

2 Lower limb orthoses

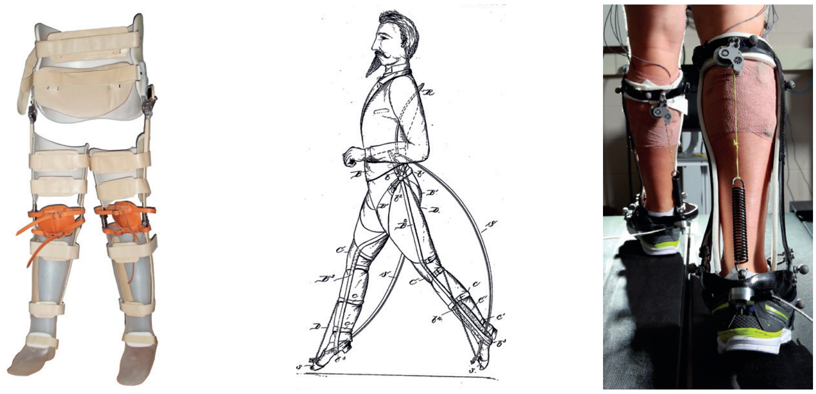
2.1 Lower limb orthoses for rehabilitation and assistance

With the latest progress in the field of assistive robotics, a variety of lower limb orthotic devices has arisen. These orthoses can potentially augment capabilities of healthy users or assist legged locomotion to people suffering from various factors affecting gait (see sections 1.4.1 and 1.4.2). Evidently not all of these devices are relevant for assisting people with disabilities but the methods implemented in some designs for completely different usages can possible be applicable to assistance for elderly. A relatively broad (but not comprehensive) selection is therefore presented here.
The orthotic devices can be organized following the way they act on the user. Their capabilities in terms of motor/driving actions are therefore used to structure this section.
In the first of the following subsections we present the passive devices which do not rely on extra power to assist the wearer. The purely passive devices rely on mechanical structures only and no electronic control occurs. With semi-passive devices, elements that cannot generate power but can still be electronically controlled are used to assist the wearer (e.g. brakes, dampers with variable viscosity, etc.).
In sub-section 2.1.2, the rehabilitation devices with body-weight support for gait training are detailed. Even though the nature of these devices makes them inappropriate for assistance in "real-world" conditions (i.e. at home, on the street, etc.), a lot of concepts are common with wearable devices. Some of these platforms are also very versatile which is ideal for testing various assistance/rehabilitation strategies.
In sub-section 2.1.3 we present the wearable devices intended to mobilize the lower limbs of people unable to move their legs by themselves. Note that here we differentiate these devices from the assistive ones. This distinction is proposed as pure mobilization does not require collaboration between the device and the user during the locomotive tasks. Indeed, in this case the user does not need to participate to the motion and is perceived by the device as an additional load. Moreover, this is not only a matter of control but also deeply depends on the mechanical design.
In the next subsection, the devices intended to augment healthy users capabilities are de-

scribed. In that case, the user is fully in charge of the movement. Here again the control of the trajectory is not shared as the device only needs to follow the user's movements. However, the nominal capabilities can be increased this way as extra load can be carried by the device and other types of perturbation can as well be rejected.
Eventually, the devices dedicated to assistance are detailed. The assistive devices are a mix of the two previous categories. In that case, a close collaboration is required between the user and the actuated orthosis so as to decrease the effort required by the former during locomotive tasks.

2.1.1 Passive orthoses

Passive devices have existed for very long enabling functional restoration of walk and walk related activities (Seymour, 2002). The most simplistic devices simply lock the articulations of disabled users thus enabling them to keep their legs straight. In the case of Spinal Cord Injury (SCI) patients for example, depending on the upper body strength, this may suffice to restore some kind of locomotion (e.g. between parallel bars, with walker, crutches,...) (Ditunno and Dittuno, 2001). Less constraining braces exists that only reduce the mobility of one articulation (usually ankle or knee). The ankle foot orthoses (AFO) are used a lot for people suffering from drop foot (very common in hemiplegic patients). The knee braces are mainly used after surgery when the Range of Motion (RoM) needs to be restrained and when out-of-plane movements are to be avoided.
Even if locking the lower limb joints may represent an improvement for some patients, natural walking requires that the articulations move. A possibility to restore a relatively natural gait with paralyzed patients is proposed with the Reciprocal Gait Orthoses (RGO). These devices couple several articulations together (typically the flexion of the hip with the extension of the knee). The number of Degrees of Freedom (DoFs) is thus reduced but a certain level of mobility is still present.
With a different philosophy, a very early patent from the 19^{th} century presented a device acting in parallel with the legs in order to facilitate walking (Yagn, 1890). However, no evidence that this device was ever built could be found. The concept seems nevertheless very similar to the ones used in some modern design.
More recently some groups have worked on articulated structures to assist walking or related activities. In (Grabowski and Herr, 2009), a device using flexible leaf to reduce the metabolic cost when hopping is presented. Actual assistance could be demonstrated for this specific task. Another possible assistive strategy is to artificially reduce the gravity effect as presented with the Gravity-Balancing Leg Orthosis (Banala et al., 2006). In (Dijk and Kooij, 2011) an approach is introduced that aims at minimizing the global mechanical work at the human joints while walking. During the stride cycle, the joints perform trajectories with phases during which energy is generated but also some others where the mechanical work is negative. If during these phases the energy is stored, it can be transferred and reused to assist gait phases that require positive work. However, even though this strategy appears very promising, the reduction of energy expenditure for the subjects could not be confirmed.



(a) (b) (c)

Figure 2.1 – Examples of passive orthoses. (a) Trunk–hip–knee–ankle–foot orthosis for enabling SCI patients to stand straight. (b) Yagn's early concept of a lower limb orthosis (Yagn, 1890). (c) Collins' light semi-passive ankle device (Collins et al., 2015).

Among the passive devices some rely on active control. The so called semi-passive devices use passive elements such as springs or dampers as a primary source of power but their action may be modulated by an active element (usually with low power consumption). The MoonWalker (Krut et al., 2010) rely on such a combination to support a part of the user's body-weight. For the same purpose an exoskeleton with passive and semi-passive joints (i.e. a variable damper at the knee) was introduced by (Walsh et al., 2007). The energy cost was demonstrated to decrease with the passive elements compared to the zero-impedance mode (when no spring and no dampers were used). However, the cost of transport was still higher than with a standard loaded backpack. The effect of the structure therefore exhibits negative aspects which must be taken into account (see 6.2). In (Dollar and Herr, 2008), a quasi-passive knee exoskeleton uses a spring in parallel with the knee to assist running. The spring can be temporarily decoupled by the action of an actuator, a feature that is used during the swing phase of the gait in order to allow the knee to freely flex. Finally, in (Collins et al., 2015), a very light semi-passive device assisting the ankle demonstrated that reducing the metabolic energy consumption was possible even without inputting any additional driving energy.

2.1.2 Gait training orthoses with body-weight support

Rehabilitation of most diseases affecting gait requires a large number of training sessions. Among the different exercises the mobilization training is of major importance in particular with SCI patients. Until a few years ago, this mobilization was performed by physiotherapists only. This task is exhausting (in particular when the patients are spastic) and often requires more than one person to perform a correct movement. With the most recent advances in robotics, rehabilitation robots have logically been developed to assist the therapists during these sessions. Precise and quantifiable mobilization is therefore possible. With the the gait training orthoses, robotic therapy went one step further towards gait rehabilitation. These devices provide verticalized mobilization thus mimicking natural walking. In addition to their mobilization capabilities they can be used to implement various gait assistance strategies.

Treadmill-based gait training orthoses

The most used rehabilitation treadmill-based gait training orthosis is the Lokomat (Hocoma, *Switzerland*) (Jezernik et al., 2003). The Lokomat enables the mobilization of the the hip and the knee in the sagittal plane while the ankle is attached on a passive joint. During the mobilization sessions, the legs of the user are guided by the actuated joints and the feet move with the treadmill during their respective stance phases. The pelvis is guided on the vertical axis and restrained in the other directions[1]. The body-weight is supported by means of a harness and a spring system. A lot of clinical evaluations have been performed with the Lokomat and many different strategies have been developed and implemented on it.
Other similar treadmill-based gait training orthoses have been proposed by different research groups such as the ALEX (Banala et al., 2007), or the LOPES (Ekkelenkamp and Veneman, 2005). Interesting results were obtained with these devices as for example the assistive strategy proposed in (Lenzi et al., 2013).

Overground gait training orthoses

Overground gait training orthoses enable the patients to train in more realistic conditions than the treadmill-based devices. Actual locomotion is then achieved and it is believed that patient motivation can be increased this way. In the existing devices a mobile frame is used to follow the user and to support his body-weight. Only two examples of such devices were found in the literature.
The first one is the WalkTrainer (Stauffer et al., 2009) designed at the Laboratory of Robotic Systems (LSRO, EPFL) and now commercialized by Swortec SA (Monthey, *Switzerland*). The WalkTrainer has two leg orthoses which are actuated in the sagittal plane (hip, knee and ankle). By design, the lateral stiffness is high but the hip adduction/abduction is allowed. The WalkTrainer is also able to actively support the body-weight of the user. An important feature of this device is the pelvis structure. The user's pelvis is guided by a 6 DoFs mechanism as its movement plays a key role in the rehabilitation process. To our knowledge, it is the only device capable of performing such a mobilization. Another specificity of the WalkTrainer is the use of the Functional Electrical Stimulation (FES). By means of a closed-loop control (the force applied by the subject on the orthosis is measured), the patient participates to the locomotive task with his own muscles (even if he is not able to control them by himself).
The second overground gait training orthoses is the CORBYS (Slavnic et al., 2014) which is composed of a holonomic platform and two orthoses. The orthoses have three DoFs at the hip (flexion/extension and adduction/abduction are active and internal/external rotation is passive), one DoF at the knee (active flexion/extension) and two DoFs at the ankle (flexion/ex-

[1] Newest versions of the Lokomat also enable lateral translation and transverse rotation of the pelvis (FreeD Module, see www.hocoma.ch).

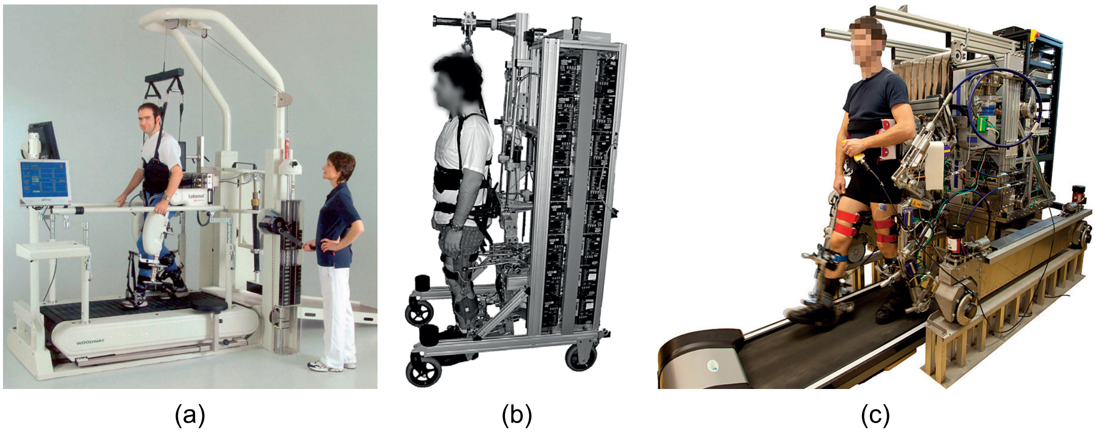


Figure 2.2 – Examples of gait training orthoses with body-weight support. (a) Lokomat from Hocoma (www.hocoma.ch). (b) WalkTrainer. Prototype version from LSRO (Stauffer et al., 2009). (c) CORBYS used in combination with a treadmill (Slavnic et al., 2014).

tension is active and eversion/inversion is passive). These features were implemented so as to enable physiological movements (e.g. turning) that other devices cannot induce.

2.1.3 Wearable mobilization orthoses

In order to restore the ability to walk to people unable to move voluntarily their legs, a device capable of generating the trajectories by itself is required. The wearable mobilization orthoses are thus used to replace the driving power that is missing with impaired users. The main distinction with the overground gait training orthoses is that the body-weight is not supported by an external frame. The balance becomes therefore a critical point that needs to be considered with attention.

Orthoses with balance control

In order to manage bipedal balance in a real-world environment a relatively complex articulated structure is required. A mechanism that acts only in the sagittal plane is *a priori* not able to ensure proper lateral balance[2]. Consequently, the devices that are able to manage this point usually possess more DoFs.
The REX from REX Bionics (Auckland, *New Zealand*) is a commercial device that provides mobilization and balance. It possesses three DoFs at the hip, one DoF at the knee and 2 DoFs at the ankle. All of them are active, which results in a relatively bulky design. This heavy structure is used to ensure safe locomotion by always keeping the center of pressure inside the support surface. The proposed gait is consequently slow and non-physiological.

[2]The lateral stabilization is theoretically possible with a devices acting in the sagittal plane only as rotation in the frontal plane can be produced with a differential action of the two legs. However, in practice this appears extremely difficult.

The Mindwalker (Wang et al., 2013) possesses five DoFs per leg, two of which are passive. The active DoFs are located at the hip (flexion/extension and abduction/adduction) and at the knee (flexion/extension). The ankle has one DoF (dorsi/plantar flexion) which is passive. The third DoF of the hip is also passive (internal/external rotation). This configuration seems to be sufficient to provide balance control. Nevertheless, this result was obtained with one unique able-bodied subject and deeper investigation is therefore required.

Orthoses without balance control

If the mobilization orthosis is not able to provide proper balance, additional means are required to increase the support base. Usually canes, crutches or walkers are used. The device is then responsible only for generating the desired trajectories. The main advantages are that the structure can be made simpler and lighter and the number of actuators can be diminished. This probably explains why this category is the most represented among the commercially available devices.
The two main commercially available mobilization orthoses without balance control are the ReWalk from ReWalk Robotics (Yokneam Ilit, *Israel*) and the Ekso (previously called eLegs) from Ekso Bionics (California, *USA*). These two devices have a very similar structure. The hips and the knees can be actively moved in the sagittal plane while the ankles are passive (spring loaded).
Parker Hannifin (Ohio, *USA*) recently presented the Indego, an exoskeleton comparable to the ReWalk and the Ekso. The Indego (previously called Vanderbilt Exoskeleton) was developed at the Vanderbilt University (Quintero et al., 2012). In 2011 when it was still a research prototype, the orthosis demonstrated its abilities to re-enable a SCI patient with a complete T10 lesion to perform locomotive tasks autonomously (Quintero et al., 2011a). A feasibility study was also conducted with FES that successfully demonstrated the possibility to use the human muscles to participate in the locomotive task (Quintero et al., 2010). Note that in this case the use of FES is completely justified for two main reasons. First, as the device is in charge both of the mobilization and of the electrical stimulation, it can be guaranteed that the two of them will work in a well coordinated manner. Second, FES imposes active participation of the patients and stimulates neural pathways which is believed to be beneficial for the rehabilitation process.

2.1.4 Wearable orthoses for performance augmentation

The wearable orthoses for performance augmentation are devices that follow the movement of their user and reject partially or completely the perturbations from the environment. The device should then ideally feel transparent to the user which implies several important points. The mobility and the "workspace" of the human wearer should not be modified. The orthoses for performance augmentation needs therefore to have the same number of DoFs as the user and the same RoM at the different articulations. Note here that this does not mean that all of these DoFs must be active. Then, the dynamic capabilities of the device needs to be

(a) (b) (c)

Figure 2.3 – Examples of wearable mobilization orthoses. (a) REX from REX Bionics (www.rexbionics.com). (b) ReWalk from ReWalk Robotics (rewalk.com). (c) Ekso from Ekso Bionics (intl.eksobionics.com).

high enough if the user wants to perform rapid movements. Eventually, the device should carry its own load in order to prevent any (high) forces from being transferred to the user. Under-actuated devices should also be designed in a way that the inertial forces are not too constraining for the user.

Several devices developed for military applications have demonstrated impressive performances. The most famous exoskeletons for performance augmentation are the BLEEX (Kazerooni et al., 2006a) and the HULC, both from Berkeley Robotics and Human Engineering Laboratory (California, *USA*). The BLEEX possesses eight active DoFs (four per leg: two at the hip, one at the knee and one at the ankle) and seven passive DoFs (one for the torso rotation and three per leg: one for the foot flexion, two at the ankle). The HULC has only one active DoF per leg located at the knee and nevertheless seems to have impressive speed and load capacity.

Another exoskeleton, HERCULES from RB3D (Moneteau, *France*), was initially developed with military motivations. Less controversial applications are now considered for this device such as help for the workers on construction sites. Even though little information is available on this device its performances appear compelling thanks to two active DoFs in the sagittal plane driven by a very well-designed screw and cable transmission.

2.1.5 Wearable assistive orthoses

The wearable assistive orthoses are devices that follow the limbs of the user but also participate actively to the movement. They are thus a combination of the mobilization devices and the performance augmentation exoskeletons. The major difficulty with this type of device is that a close collaboration with the user is required. Neither the device nor the user is fully in charge of the locomotive task. In order to have an effective assistance, the human and the machine must be coordinated and work together. Moreover, the orthosis must be as little intrusive as

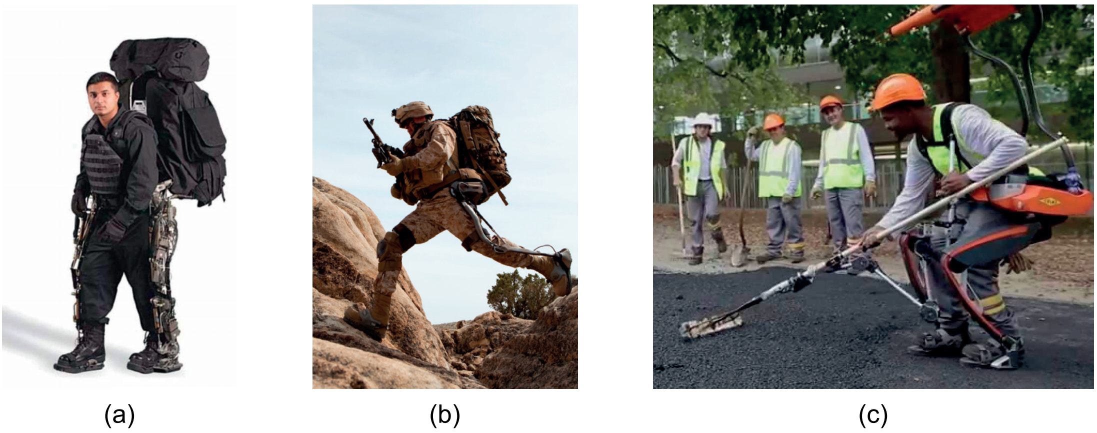


Figure 2.4 – Examples of wearable orthoses for performance augmentation. (a) BLEEX from the Berkeley Robotics and Human Engineering Laboratory (Kazerooni et al., 2006a). (b) HULC also from the Berkeley Robotics and Human Engineering Laboratory (Kazerooni et al., 2006a). (c) Modified version of the HERCULES used in Switzerland by Colas company (www.rb3d.com).

possible. This means that hindering the movements should be avoided at the most.

A relatively effective way to interface a wearable assistive orthosis with a subject was proposed with the HAL from Cyberdyne (Tsukuba, Japan) which was initially developed at the University of Tsukuba (Kawamoto and Sankai, 2002). The HAL measures the motor action potential on the skin in order to estimate the user's intention and move accordingly. Even though the actuation of the HAL is very similar to the ReWalk or to the Ekso, the way it acts on the wearer is very different. Complete SCI patients cannot use the HAL as it does not provide pure mobilization. Instead it requires that the user is able to, at least, send nervous signals to his muscles. In that case, the sensing and the control strategy fully define the way the orthosis acts on the user, which is rarely the case as the mechanical design is usually crucial. This control strategy will be further discussed in section 2.2.

Despite its proven usefulness, electromyography (EMG) is not always desired as it is relatively invasive (electrodes need to be pasted on the skin). Furthermore, the quality of the signal may degrade over time (e.g. when sweating). Kong and Jeon proposed an alternative to EMG with EXPOS (Kong and Jeon, 2006). Here again the intention is detected at the muscle level. In their design air chambers positioned on the muscles and pressure sensors are used to measure the muscle contraction. The EXPOS possesses another peculiarity that addresses one of the key points of assistive orthoses. We remind here that the assistive devices need to be as light as possible in order not to impose high inertial or gravitational forces that may hinder the user's natural movements. In order to reduce the amount of heavy parts worn and carried by the user, the actuation, the power and the control units of the EXPOS are remotely located in a "caster walker" (i.e. a heavy box mounted on wheels that the user can use as a walker). The driving torque is then transferred to the different joints by means of cables and pulleys. EXPOS has four active joints (the two hips and the two knee in the sagittal plane) and four additional passive joints per leg at the thigh and at the ankle level. Kong and Jeon reported that their design was performing "fairly well" but no publication more recent than 2006 could be found.

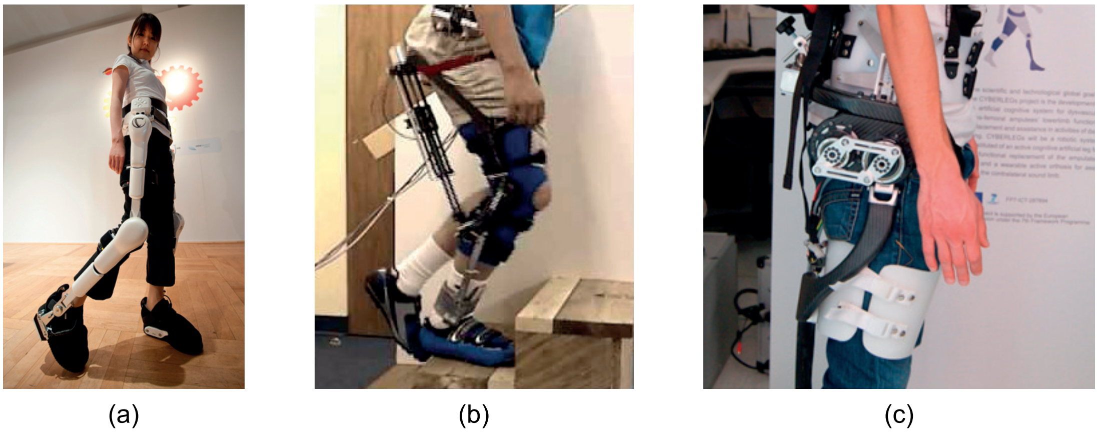


Figure 2.5 – Examples of wearable assistive orthoses. (a) HAL from Cyberdyne (www.cyberdyne.jp). (b) RoboKnee partial device for the knee only (Pratt et al., 2004). (c) Vitiello's group hip orthosis (Giovacchini et al., 2013).

Another method to limit the invasiveness of a wearable device is to assist only partially the limb. Several devices using this approach were developed for the ankle, the knee as well as for the hip.
As presented in subsection 2.1.1, ankle orthoses can potentially decrease the metabolic cost during walking. Before this was demonstrated with a semi-passive device, similar result were obtained with active device as for example in (Cain et al., 2007). A particularity of the device that was used is that the actuator is a pneumatic artificial muscle. Therefore it can only assist the plantar flexion as these actuators can only provide force when they contract. Also worth mentioning, an ankle orthosis using Series Elastic Actuators (SEA) successfully demonstrated clinical benefits for the treatment of drop-foot gait (Blaya and Herr, 2004).
SEA have similarly been used in an assistive device for the knee with the RoboKnee (Pratt et al., 2004). This device was presented by its authors as a performance augmentation exoskeleton but it really acts on the user as an assistance on one particular articulation. It therefore does not reject directly the perturbations from the environment while following the user's movement. In our terminology the RoboKnee is then an assistive device even though it was developed to enable a healthy user to carry a significant load in a backpack, which relates more to performance augmentation.
To our knowledge, the first assistive device for the hip was presented by Honda as a non-commercial prototype. Very little information is available on this device but in (Shimada et al., 2008), it was used to study the influence of an assistance[3] on elderly subjects. More recently, Prof. Vitiello's group presented a hip orthosis (Giovacchini et al., 2013). The capabilities of this device in terms of transparency appear very promising thanks to the SEA. This characteristics is important in order not to hinder the natural movements of the user (see 2.2). Preliminary tests on assistance have been performed but no conclusive results have been published yet.

[3]Unfortunately only vaguely defined.

2.2 Control strategies for actuated orthoses

In this section we present the state-of-the-art techniques for controlling lower limb actuated orthoses in the context of locomotive Activity of Daily Living (ADL). The challenges regarding the way the orthotic devices are controlled in concert with the user's impaired sensory-motor control system is at least as exigent as the hardware engineering ones (i.e. light and robust articulated structure, actuation, energy storage, miniaturized sensing, etc.). This means that here, only the active wearable devices for impaired users are addressed. This excludes the passive devices and the non-wearable systems. The devices for capabilities augmentation are only briefly mentioned when it appears relevant in specific cases. The challenge-based strategies which make the movements more difficult to perform are not presented either. Even if these demonstrated their abilities to provoke motor plasticity, they are not relevant in the context of assistance for daily life activities. In substance, this section is an adaptation of the paper on *Control strategies for active lower extremity prosthetics and orthotics: a review* (Tucker et al., 2015). For this reason, some of the explained concepts come from prostheses as they also fully apply for orthotic devices.

2.2.1 Generalized control framework

In order to structure the classification of the various control approaches for orthotic devices, we propose the generalized framework of Fig. 2.6. This framework was inspired by and extended from that of Varol et al. (Varol et al., 2010). This diagram depicts the physical interaction and signal-level feedback loops between a user, the environment and a powered assistive device. The main components of this figure are a hierarchical control structure, the user of the powered device, the device itself and the environment through which the user ambulates. A safety layer has been included in order to emphasized the importance of this consideration in the context of human-robot interactions, especially considering the amount of power that the considered devices are able to generate. On Fig. 2.6, each subsystem is defined by a set of physical and signal-level inputs, by a set of processes and by its outputs. The processes operate on the inputs to control the exchange of power through each subsystem. The outputs transmit the power and the signals to the connected subsystems.
The terrain features and surface conditions of the environment (environment state) defines the movements that can be performed. The information from the environment may then possibly be used by the controller to optimally adapt the locomotion strategy. Therefore it acts on the user, on the device and conceivably on the controller.
The motion intentions are initiated by the user. Therefore his physiological state and desire must be interpreted. The user's state is defined by the positions and velocities of his body parts and by the interactions with the environment and the device. The motion intention estimation requires that the user's state is properly sensed. The high-level controller must perceive the user's locomotive intent in order to properly identify the activity mode (e.g. standing, level walking, stair climbing/descending, etc.). With a direct volitional control the user is enabled to directly manipulate the device positions, velocities and torques. The volitional control can

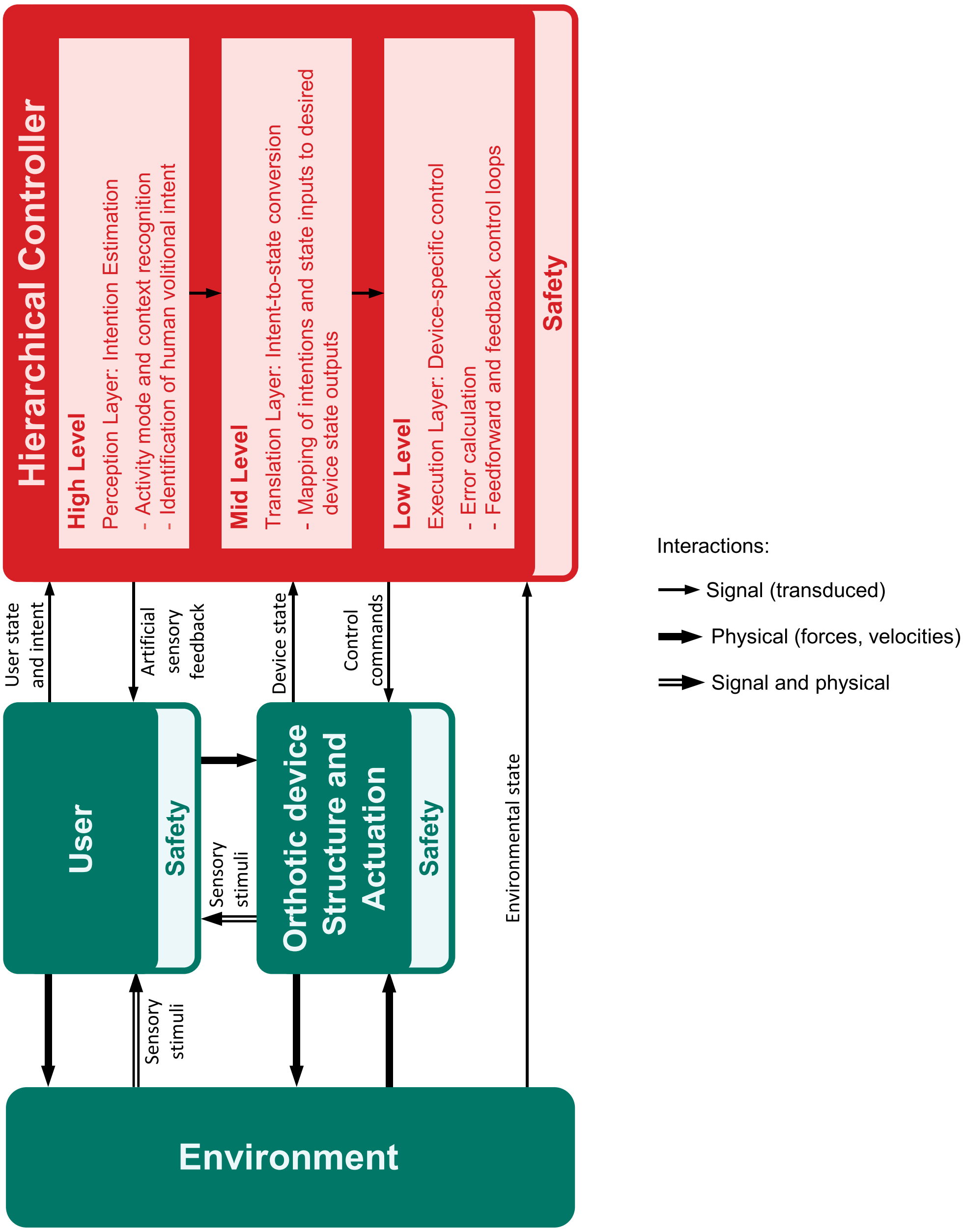


Figure 2.6 – Generalized control framework for active lower limb orthoses. The proposed framework illustrates the physical and signal-level interactions between a powered orthotic device, a user, and his environment. The arrows indicate the exchange of power and information between the various components. A hierarchical control structure is implemented. The estimation of the user's locomotive intent is taking place at the high level. The translation of the user's intent to a desired device state is managed by the mid level controller, and a device-specific controller is responsible for realizing the desired device state at the low level. Safety mechanisms must be present in all aspects of the orthosis design. Adapted from (Varol et al., 2010).

be combined with the intention detection. In that case it can be used to modulate the device behavior within a particular activity.
The mid-level controller translates the user's intentions from the high-level to the device states for the low-level to track. The gait cycle is determine at this level and the desired control law is applied. Position/velocity, impedance or admittance are the typical states which are sent to the low-level controller. The low-level then computes the error with respect to the current state and sends commands to the actuator(s) in order to reduce it. Typically, feedback and/or feed-forward control is used to achieve the desired behavior. The kinematic and kinetic properties of the device are taken into account at this level.
Finally, the device is actuated to executes the commands of the low-level controller in order to act on the user and on the environment.
In the following subsections, the environmental interactions and the controller strategies will be discussed with regard to the roles that they play in the proposed generalized control architecture. The safety aspects will addressed in subsection 2.2.4.

2.2.2 Environmental interactions

The environment provides the reaction forces enabling balance, support, and propulsion. These forces are a function of the ground contact surface condition and of the slope of the terrain. Other forces arise due to the physical properties of the environment, such as gravity or fluid dynamic drag. Obstacles may as well be present in the environment. They may therefore impede the movements in a particular direction, thus forcing the user to circumnavigate or to perform a compensatory motion to negotiate them. Each of these environmental properties has a major influence on the stability, the balance, and the energy consumption of the user and of the device (Riener et al., 2002). They should thus be considered in the overall control scheme. The state of the environment can be indirectly deduced based on the states of the user and of the device or directly estimated using dedicated sensors. The information about the environment can be used to define a control policy over a time window of several steps or as information that can influence reactions within the current step.

Implicit environmental sensing

Typical environmental features may be extracted implicitly from the states of the user and of the device at various instants of the gait cycle. For example, the slope can be estimated using an accelerometer mounted on the foot as presented in (Li and Hsiao-Wecksler, 2013). When the heel and the toe of the foot are in contact with the ground surface, if there is no slip, the acceleration vector matches that of the gravity and the orientation can be inferred. An Inertial Measurement Unit (IMU) may also be used to detect the elevation. This can be used for example to detect successive steps during stair ascending/descending (Li and Hsiao-Wecksler, 2013)(Li et al., 2009).

Explicit environmental sensing

Very few examples describing orthotic devices including explicit environment sensing were found. This may be attributable to the fact that most of the devices so far are dedicated to one or few activities. Moreover, they are usually operated in a well-defined and controlled environment. It is however expected that sensing the environmental state will gain higher priority in order to bring the orthotic devices into real-world context. Some wearable devices already exist that can estimate the height and slope of a terrain. Zhang et al. (Zhang et al., 2011) presented such a system on an able-bodied user. It is based on a body-worn laser distance sensor (attached to the waist) and IMUs fixed to the limbs. In the context of prosthetic devices, Scandaroli et al. presented a method using gyroscopes and Infrared (IR) sensors for estimating the slope and elevation of the foot above the ground (Scandaroli et al., 2009). Examples from other field of research exists, for example with brain-controlled wheelchairs. By using an array of sonar sensors and digital video cameras it is possible to detect obstacles and to use this information in a shared control for example to circumvent obstacles (Carlson and del Millan, 2013).

Environmental context

Knowledge about the environment through which the user moves is crucial as it constrains the likelihood of encountering specific features. Moreover, the properties remain usually quasi-constant over time, unless the environment is unstructured (e.g. a rocky hiking trail, a child's messy room) or has variable surface conditions (e.g. snow, sand, loose gravel). The contextual information could typically be used when the user is inside a modern public building. In that context, the floor is usually flat and level, stairs are regularly spaced, and accessibility ramps will have a slope that is known in advance. Therefore if a device is able to localized itself in such an environment, the high-level decisions can be weighted depending on the specific current context and the mid-level controller can be optimized for the most likely terrain.

2.2.3 Control strategies

The controller for an orthotic device can be divided into three parts (see Hierarchical Controller on Fig. 2.6). The intention of the user is detected by the high-level controller which interprets the signal from the user, the environment and the device. The mid-level controller then translates this information (i.e. the user's motion intention) to a desired output state for the device. The low-level controller eventually executes the device-specific control loop to obtain the desired trajectory. Note here that the term trajectory may refer to something different than simply position (e.g. trajectory of force).

High-level controller

The purpose of the high-level controller is to perceive the locomotive intent of the user through a combination of activity mode detection and direct volitional control. Depending on the user's pathology, his ability to generate, transmit, and execute appropriate commands may be impaired at some level. Therefore, the proportion of what is executed by the user or by the device may vary. Once the high-level intentions are provided, the device should be responsible for the execution of the mid- and low-level tasks. The purpose of this shared control approach is to limit the cognitive burden imposed on the user (Millan et al., 2004).
Switching between different locomotive activities should ideally be performed without any conscious input from the user. In addition to this activity mode recognition, a direct volitional control can be coupled. In that case the user is given the ability to modulate the device behavior within a particular activity (Simon et al., 2013). Note that a pure volitional control (without activity mode recognition) is also feasible.

Activity mode recognition is what enables the high-level controller to switch between the mid-level controllers. Each of these controllers is appropriate for one particular task (e.g. standing, walking, stair climbing,...). Most of these activities have a cyclic and relatively long-lasting nature which enables to use automated pattern recognition methods for classification. The important considerations for choosing a classifier are the number of activities that can be discriminated, the procedure required for training, the error rate, the signals that are required as input and the classification latency. The latter is the time that is required to reach a decision. Huang et al. defined the term *critical time* to describe the maximal time to reach a decision for the classifier in order to guarantee a correct and safe transition between modes (Huang et al., 2011). The critical time is of major importance when switching between activities which differ considerably. For example, a late transition between level walking and stair climbing may cause a loss of balance or even a fall. Zhang et al. use the term *critical error* to describe any error that results in the user's impression of unstable balance (Zhang et al., 2012). This highlights the importance of the subjective feeling of safety which must complement the actual balance.
In the following paragraphs we present the different types of activity mode recognition classifiers which can be used in motorized orthotic devices.
Heuristic rule-based classifiers represent an effective method for identifying mode transitions. Many examples of finite state machines (Quintero et al., 2011b) and decision trees (Li and Hsiao-Wecksler, 2013)(Kawamoto et al., 2003a)(Novak et al., 2013) exist. In these methods, a fixed set of rules is defined that indicates the transition from one gait mode to another. These rules are based on the sensed state of the user, the device or the environment. In (Li and Hsiao-Wecksler, 2013) a transition from level walking to stair ascent is detected by a change in the elevation of the foot from the beginning of one step to the next. In (Kawamoto et al., 2003a), the Ground Reaction Forces (GRF) and the position of the hip and knee joints are used to identify sitting, standing and walking. Note that while the rules are selected heuristically in that case, the used criteria may either be manually selected (Kawamoto et al., 2003a) or

determined through analytical means (Li and Hsiao-Wecksler, 2013)(Novak et al., 2013). In order to avoid inappropriate mode switching back and forth, thresholds with hysteresis may be used (Sup et al., 2011).
Automated pattern recognition is based on machine learning techniques and has generated a variety of classifiers (Huang et al., 2009)(Kilicarslan et al., 2013). The term "automated" means here that the generation of the classification boundaries during the training is continuously learned. The classification is therefore automatic which is not the case with the rule-based classifiers. After a period of supervised training has been completed, the classifier can be used to define new classes to newly observed sets of data. The main benefit of these methods compared to the rule-based classifiers is that data from a large number of sensors can be input to the classifier. The controller may therefore be more accurate and robust as the decisions are computed based on more information. Note that a manual boundary identification from a large variety of sensors would also be difficult. As explained, automated pattern recognition requires training data in order to properly classify the different activities. The training data need to have enough variability in order to keep suitable performances in real-world conditions. Moreover, the data set for training should come from the user himself which is sometimes difficult or even impossible if the user is too severely impaired.

Direct volitional control enables the user to modulate the device state within a recognized mode. This can be particularly important when the activity is irregular or non-cyclic. Indeed, situations like walking in a crowd or on a rough terrain or simply repositioning one's legs during non-locomotive activities may require a direct control from the user. Electromyography (EMG) has been largely used for that purpose (Kawamoto et al., 2003b)(Fleischer and Hommel, 2008)(Ferris et al., 2006)(Karavas et al., 2013) as it is a very direct way to detect voluntary intentions.

Mid-level controller

The mid-level controller is responsible for converting the estimated locomotive intent output from the high-level to a desired device state that the low-level controller can use. Usually several mid-level laws can be used depending on the output of the activity mode recognition. The coordination between the different joints of the device is also managed by the mid-level controller. The inputs of the mid-level controller are the states of the user, the environment and the device.
Two main types of mid-level controllers can be distinguished. The first ones (phase-based) depend on the gait phase while the second do not (non-phase-based).

Phase-based controllers are based on clearly identified events such as for example the heel strike or the toe-off. After the detection of such an event, a set of actions is performed following a programmed sequence. The timing is then very important which implies that the regularity of the steady-state stepping period is crucial. These controllers are therefore not adapted to

deal with irregular situations or to react to unexpected events.
A very common mid-level controller is based on finite states (see Fig. 2.7) which are a decomposition of the gait in distinct phases. Many examples can be found in the literature (Shorter et al., 2013)(Li et al., 2011)(Li and Hsiao-Wecksler, 2013)(Kawamoto et al., 2003a)(Boehler et al., 2008)(Murray and Goldfarb, 2012). The finite-state controllers (FSC) typically set different control laws depending on well delimited phases. These phases are then repeated cyclically over consecutive strides. It appears then obvious that a different FSC is required for each activity mode (defined by the high-level controller). Moreover, as the gait patterns may vary from one individual to another, it is necessary to tune the controller parameters to optimize the user's comfort and the efficiency.
Orthotic devices for people with pathologies affecting only one side can be operated with the so called echo control. This method uses the position trajectories of a sound limb in order to replay them on the other limb. Assistance is then provided to the affected side based on time delay and scaling. Half a period delay is typically used during steady-state walking. For activities such as standing up from a sitting position, no phase shift is required.

Non-phase-based controllers do not use specific events of the gait cycle and directly use instantaneous inputs.
An example of such a method is the force-feedback controller which measures the interaction force between the user and the device and acts to reduce it. When the interaction is completely compensated for, transparency or zero-force is achieved. The transparency that can be achieved with a device can also be used as a performance metrics. Indeed the transparency allows straightforwardly to define the maximal dynamical performances that can be achieved with a device. Force-feedback is also used in performance augmentation tasks (Bogue, 2009). In that case it enables the user to maintain his agility while having his force amplified.
The Complementary Limb Motion Estimation (CLME) (Vallery et al., 2009) is another non-phase-based controller. Similarly to the echo control, the movement of a sound limb is used to infer motion on an affected limb. Contrarily to the echo control, the CLME complements residual body motion without delay. The intended motion of the affected limb is inferred from the sound articulations and mapped to a reference trajectory. This is achievable as a strong inter-joint coordination is exhibited during human motion (in particular during walking). The mapping is derived through regression of physiological gait recordings of healthy subjects. The training data come therefore from a different subject. Even so, this controller demonstrated its effectiveness to restore functional gait patterns.

Low-level controller

The low-level controller is responsible for driving the actuators in order to reduce the error between the device current and desired states. This level of controller fully depends on the device. It relies on a combination of feed-forward, compensation and/or feedback control loops. For the first two, a model of the device is used to reduce the undesired interaction forces

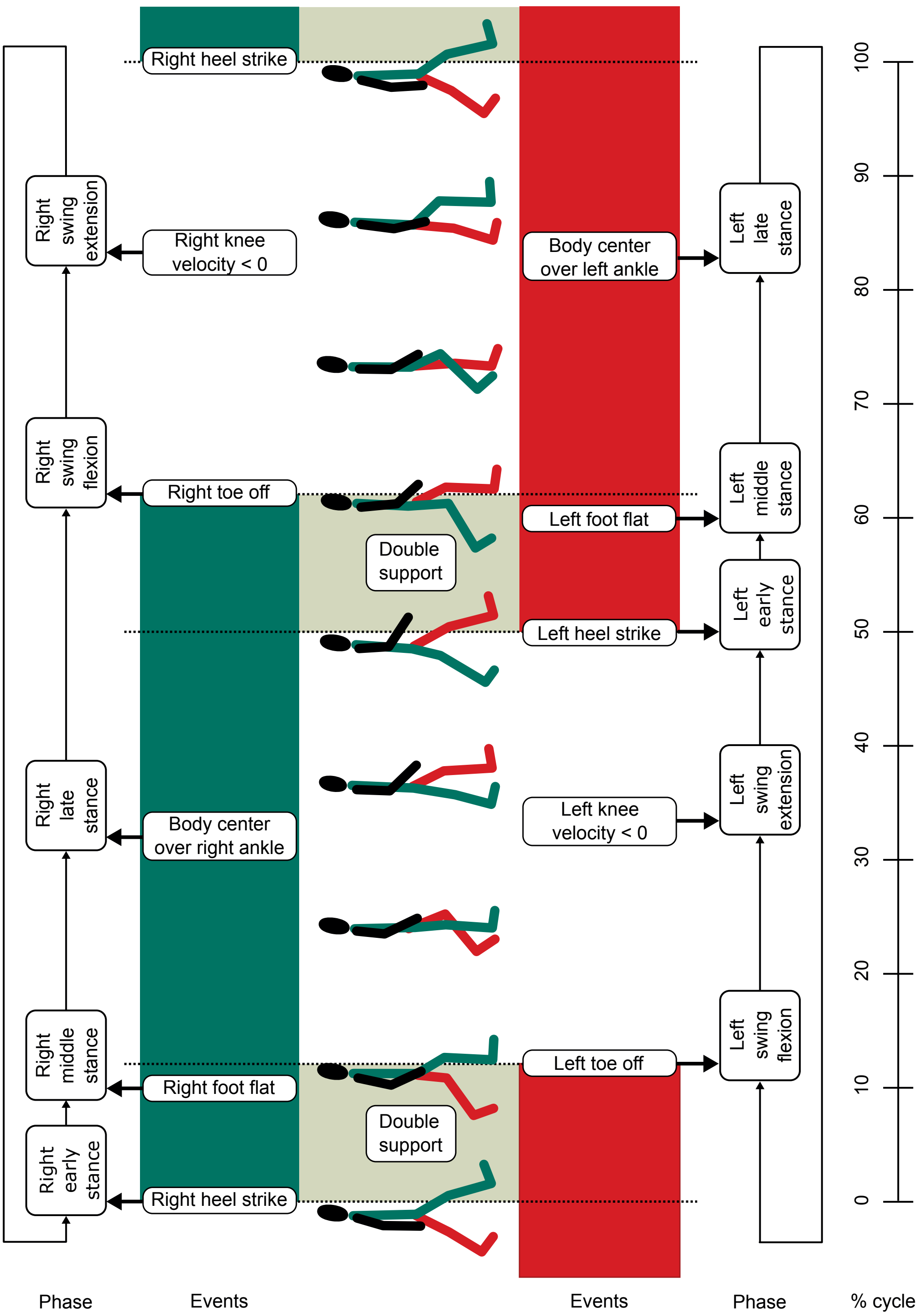


Figure 2.7 – Finite-state decomposition of level human gait. Steady-state locomotion can be represented as a periodic sequence of states (or phases). The transitions between the states are triggered by events within the gait cycle. In this example for the knee joint, stance has been divided into three states. The early and middle stance are initiated by ground contact events at the heel and toe of the foot. Late stance is triggered when the user's center of mass is estimated to be over the ankle. Swing flexion begins as the toe of the foot leaves the ground, and swing extension begins as the knee velocity is sensed to be less than zero.

due to inertia, mass, and friction. The compensation relies on the information coming from the sensors while the feed-forward uses on the command to anticipate the dynamic effects. In turn, the feedback controller compares the desired state with the estimated state from sensors information. The performance of the system is then limited by modeling inaccuracies, sensor noise, signal quantization, discrete sampling effects and control loop execution time.

2.2.4 Safety mechanisms

The active orthoses being able to provide significant torques/forces (comparable to human capabilities if not greater) they are potentially hazardous. Safety mechanisms are therefore vital so as to guarantee that nothing severe may happen to the human, the environment or to the device itself. Concerning the former, not only the physical risks are to be taken into account. Emotional and psychological effects are also important and requires consideration (Rubenstein, 2006).

The main risk during legged locomotion is the fall, which may be induced by tripping, slipping, etc. The orthotic device is possibly an additional source of such incidents for example in case of control failure. Encountering unexpected terrain may also provoke loss of balance. The risk must then be evaluated according to its severity, its probability of occurrence and whether it is detectable.

In order to prevent dangerous situations, several approaches can be implemented. First, passive mechanisms can be used. Mechanical stops which guarantee that the orthosis has smaller ranges of motion than the user's is one of those. Torque and force limiters are also conceivable to make sure that the user will not endure to large efforts. All sorts of fuses (mechanical and electrical) and switches (operated by the user himself) can similarly be used to increase the safety. Then, active mechanisms are also desirable. Redundant sensing appears compulsory in order to avoid failures due to incorrect measurements. Similarly to the mechanical stops method, the range of motion can be limited by the controller. Other more complex strategies are also implementable at this level. For example, if a failure of any kind is detected, the controller may switch to a "safe behavior". The safe behavior depends on the device and on its utilization. A wearable device for assistance can typically switch to a transparent mode as its user should be able to move by himself. This cannot be the case with a mobilization device with which a locking of the joints appears more appropriate. In any case, the users should be alerted if a failure is detected. As a last resort in case of fall, a radical measure implemented by the commercialized ReWalk orthosis is a wearable airbag system that is deployed either manually by the user or autonomously using built-in sensors.

Several studies have discussed about the consequences of failures. The following cases have been addressed: misinterpretation of the user's intent or invalid sensors inputs (Zhang et al., 2015), unstable interaction with the user and the environment (Fleischer and Hommel, 2008)(Vallery et al., 2008), actuation failure (Kazerooni et al., 2006b) and controller failure due to software bugs or overloaded computational resources (Fleischer and Hommel, 2008).

In 2014, the International Standards Organization (ISO) published the standard 13482 in which definitions, design guidelines and safety requirement for personal care robots are expressed.

This standard is not explicitly made for medical devices (as most of the orthotic devices are considered). However it does apply to wearable suits and exoskeletons for physical assistance. Other methods exist that help identifying potential causes of injury or damage. We name here the Failure Modes and Effects Analysis (FMEA), HAZard and OPerability (HAZOP) and HAZard ANalysis (HAZAN) studies, and Fault Tree Analysis (FTA).
Before the commercialization of orthotic devices used in activities of daily life, the developers must first establish trust with regulatory bodies, funding agencies, insurance companies and ends users. Clinical trials are therefore compulsory. As example, the ReWalk was the first RDA-approved powered orthosis to be marketed for personal use via the "de novo" classification process. This classification is used for novel devices which underwent extensive performance testing and clinical studies. The use of this kind of classification, however, indicates that there is still a gap to be filled in the regulation.

2.3 Discussion

Many devices exist that are used to help their users during locomotive tasks in different ways. Both the hardware and the control are important to achieve the desired results. Therefore they must be considered and designed in consideration one with another.
For the most impaired users, functional restoration of basic locomotion or even rehabilitation of basic movements is addressed. With the latest progress in the field of mobilization orthoses, some of these patients (the ones who have a good upper body strength) are even given the possibility to autonomously stand up from a chair, walk and even climb/descend stairs in a quite natural way. With such devices, a combination of movements locking and joints mobilization is performed in order to execute the user's intention. A more comprehensive mobilization is also possible (e.g. with devices like the REX) but the gait is no more physiological that way as the increase of complexity implies additional weight. The dynamical performances of these complex structures are therefore modest.
In the case of people having their initial strength but who need to perform tiring tasks, a support is proposed in order to increase their nominal performances. The performance augmentation devices follow the movements of the user and increase their output force on the environment. As the given task may require precise trajectories, the user's operations should not be hindered by the device. Therefore the mechanical design is crucial to enable all the possible desired movements.
In this chapter (notably in section 2.1.5) we mentioned that a similar hardware may exhibit a very different behavior if appropriate sensors and control strategies are implemented. The HAL was compared to the Ekso and the ReWalk which act as mobilization devices while the orthosis from Cyberdyne is (according to our terminology) an assistive device. We remind here that this difference in behavior is only made possible by the use of Electromyography (EMG). This technique is remarkable but remains relatively invasive and not reliable for extensive periods of use. Moreover, even though the driving action of the orthosis is well integrated by the user and therefore provides assistance, the out of plane movements are still restrained and

the user can only perform pre-programmed movements.
Patients suffering from a moderate weakness and who are still able to move by themselves may benefit from the assistance of devices that work in collaboration with them. This type of devices is at the crossing between the mobilization orthoses and the performance augmentation devices. Indeed, the movements should not be restricted (at least most of them) but force should be transferred onto the user. The wearer is still the master but the slave does not simply follow her/him anymore. An appropriate assistive strategy is then required to empower a natural integration. Here again, a strong dependency between the hardware and the possible strategies that may be implemented exists.
Other techniques well adapted for assistance have been developed, which allows to dispense with the EMG. The SEA for instance were mentioned several times as they appear ideal for implementing assistive strategies on orthotic devices without increasing their invasiveness (or only slightly as these actuators are usually a bit heavier). Artificial muscle based and semi-passive ankle orthoses have also demonstrated that simple but well-designed devices may have a beneficial impact on the metabolic cost when walking. Note that this was not the case with more complex multi-joints spring-based structures. This highlights the importance of the mechanical features (weight, kinematics, RoM,...) and their possible negative impact due to additional undesired efforts.
Non-wearable gait training systems were initially developed for rehabilitation purposes in a clinical environment. They are thus not adapted to "real-world" scenarios. Nevertheless, many concepts are shared with the wearable systems and a lot can be learned both from the hardware as well as from the control strategies points of view. FES for instance was first evaluated on non-wearable systems and then used in mobilization devices. Moreover, as they offer a safe setup for testing thanks notably to the body-weight support, control strategies can be implemented on them before being tested on wearable orthoses.
The safety aspects were also addressed in this chapter as the presented devices are powerful enough to be potentially hazardous.

3 Assistance for the elderly

As presented in chapter 2, there are many solutions that provide support for walking. However, the problem of assisting a person who is capable of walking but needs assistance to gain in autonomy is not convincingly addressed. In order to better respond to the seniors' needs in terms of assistance, in this chapter we propose guidelines to develop assistive devices. In the second section, the options that were chosen are contextualized and justified. Eventually, detailed technical requirements are established.

3.1 General guidelines

The factors affecting gait in elderly may be due to aging or caused by specific pathologies. In any case, objective observations can be made in order to pinpoint the local source(s) of a gait impairment (e.g. incapacity to produce high torques at the ankle level, reduced articular Range of Motion (RoM), etc.). This can be done without necessarily knowing the origin of the problem. In 1.4, various conditions are concisely described and the mechanisms leading to the walk alterations are developed. From that, it may be noticed that the factors acting at a local level occasionally overlap. For instance, diminished ankle power capabilities is induced by a large number of pathologies. The compensation strategies adopted by the people suffering from one condition or another may then also be comparable. In the context of the limited plantar flexion induced by different pathologies, a similar reorganization of the lower limb articulations coordination is likely as they are initiated by a shared local source of impairment. In addition, the walking alteration in elderly is very often multifactorial. Part of the different strategies adopted by the seniors may then have common sources. It is also likely that some gait modifications have a positive impact on several local factors. The gait parameter that is the most commonly modified in people who have gait disorders is the progression velocity. This is easily explained by the fact that speed reduction has a significant impact on many important aspects such as: stabilization of the head where the visual and the vestibular systems are located, balance in general, articulation torques, global energy consumption, etc.

Even though the adaptation strategies have sometimes similarities, a universal solution appears unrealistic. The developments presented in this thesis therefore focus primarily on one

particular problem, namely the diminished muscular force and in particular the incapacity to generate high torques at the ankle level. The neurological problems are not directly addressed in this work as it is important, in the first instance, to ensure that the assistance is well received by people with proper control of their body. It is however envisioned that the developed devices may be later used to study other types of gait impairments.
Several different approaches can be adopted to assist a person with gait deficits. In this work we focus on assisting the articulations with a robotic device. Two main strategies can then be implemented to promote safe locomotion. The first one focuses on "repairing" the affected functions in order to restore nominal gait while the second concentrates on facilitating the compensation strategies adopted by the user. The latter may then lead to modified gait patterns (in comparison with young people or even with healthy elderly). The first technique is *a priori* more justified as it tackles the causes (or at least a higher level in the causal chain) and not the effects. Nevertheless, the compensation strategies being the result of a re-optimization adapted to the global modified capabilities, it may also be justified to promote them. Moreover, as the origin of the gait alteration is not necessarily well mastered, a direct action on the visible modification may be hazardous. In the case of peripheral neuropathy (PN) for instance, acting at the ankle level may provoke increased Ground Reaction Forces (GRF) that are known to be damageable for an insensitive foot (see 1.4.2). The difficulty is then to select the most suitable assistance for a specific condition.

3.2 Proposed solutions

The problem addressed in this work is assistance as opposed to mobilization. Pure mobilization is not adapted to the elderly suffering from muscle weaknesses as it is a hard constraint that restricts their deliberate actions. Rehabilitation (or training) aims to restore autonomy but it may not suffice. The improvement of functional walking is then crucial. The goal is thus to provide elderly people with a device that helps them with the locomotive task and that impedes as little as possible on their Activity of Daily Living (ADL). The restoration of conventional gait patterns is secondary. For ADL, the invasiveness of an assistive orthosis must be diminished to its limit. Its size and weight should therefore be reduced at the most and the interaction with the user must be smooth, comfortable and safe.
The concept that we propose to investigate is the use of **partial devices** with **minimized invasiveness**[1]. We suppose that the effects of a device that assist one articulation to perform one particular movement may have benefits above the joint it assists. This hypothesis is driven by the fact that people suffering from a weakness (typically at the ankle) are able to compensate with a reorganization of their coordination (in the case of the ankle weakness, with the hip-strategy).
The most direct benefit of an orthosis for a single articulation is that part of the invasiveness is reduced in a very simple manner. The weight is limited and no burden is induced at the joints

[1] Invasiveness should be understood in its wide sense. A device that hinders movements, that is bulky or heavy is considered invasive in our use of this term.

where the device is not present.
The invasiveness being multi-faceted, new approaches are hence required to:

- prevent the reduction of the articulations RoM or the locking of Degrees of Freedom (DoFs),
- limit the undesired dynamic effects (notably due to gravitational, inertial or frictional forces),
- intuitively work in concert with the users by providing the right amount of assistance when needed.

The main focus of the present work is given on the hip joint. This choice is justified by the fact that many compensation strategies solicit primarily this articulation (see 1.4). The elderly tend to rely more on this joint even though it is often subject to bone weaknesses and Osteoarthritis (OA). An assistive device acting at the hip level could then potentially facilitate the implementation of the compensation strategy without worsening an initial frail articulation condition.
The other articulation that requires attention is the knee as a very large part of the population suffers from OA at this joint. The developments presented in this work therefore also address this problem, but to a lesser extent.
No development have been made to assist the ankle joint. The first reason is that we intend to assist the compensation strategies and it is very uncommon to have such strategies acting at the ankle level. Usually the hip or the knee are more solicited. The second reason is that passive, semi-passive and active devices addressing this articulation in the context of assistance already exist. It appears that some of them have proven their effectiveness at assisting the ankle, notably in (Collins et al., 2015).

3.3 Technical requirements

This section is an attempt to translate the above-mentioned guidelines into objective specifications. The RoM of the different DoFs are specified and the movements that most require assistance are identified. The maximal weight and the requirement in term of dynamics, torque, power and transparency are then established.

3.3.1 Kinematics

An assistive device is *a priori* not supposed to lock any DoF. Consequently, all the DoFs described in 1.3.1 and in particular on Fig. 1.2 need to be preserved.
The movement amplitudes in elderly is not dramatically different than that of younger people except conceivably for the hip extension (Roach and Miles, 1991). The RoM reported in Table 1.1 can then be used as reference. Approaching the maximal amplitudes with a device

potentially able to transmit high torques may however be hazardous. For safety reasons, the active DoFs should therefore have mechanical stops that are within the standard RoM (see Table 1.1). For the hip extension, which is one of the most critical, it may even be necessary to increase the safety margin and limit the angle at 5°.

3.3.2 Movements requiring assistance

The goal being to assist walking with an under-actuated structure[2] the selection of the movements to assist is essential.

Hip

The hip being very comparable to a spherical joint, three rotations are possible and consequently torques in all directions may be required. Due to the the relatively important thickness of the soft tissues at the thigh and because of their anisotropic behavior (Ridge and Wright, 1966) not all the torques can be efficiently transferred to the body. The internal/external rotation for instance can hardly be assisted because the soft tissues have a tendency to deform and rotate around the femur. The order of magnitude of this deformation is similar to that of normal internal/external rotation during walking. Moreover, it seems that the torque in internal/external rotation is not so commonly affected in elderly with walking disorders. On the other hand, for the flexion/extension which requires an important RoM, the compression of the soft tissues appears negligible and this movement can be assisted with less difficulties. Moreover as its RoM and its peak torques are the largest at the hip, this movement is often considered as the main action of the hip during walking. Furthermore, the "hip-strategy" often adopted by the elderly to compensate for a weak ankle requires that more torque is generated at this level (DeVita and Hortobagyi, 2000). Assisting this action therefore appears logical. The adduction/abduction is usually less important than flexion/extension but some compensation strategies rely on this movement[3]. Assisting this movement could then be interesting for some pathologies such as myelopathy (see 1.4.2) but is not implemented in the presented developments. Note that in that case again, the soft tissues deformation would not represent a major limitation.

Knee

The selection of assistance at the knee level does not represent a major challenge as the movement can, in first approximation, be considered as pure flexion/extension (see 1.3.2). The torques possibly required about the other axes can then be provided with passive structure

[2]Not all the articulations are assisted and some DoFs of the assisted articulations are passive.
[3]Often described in those cases as circumduction as it is combined with flexion/extension

which would then act as stabilization for the articulation (see 1.3.3). Indeed the knee adduction/abduction torque is present only in the stance phase when there is no motion about the antero-posterior axis. Moreover its pattern relatively accurately corresponds to that of the hip, which implies that it is a reaction torque and therefore does not need any active assistance. The rotation about the longitudinal axis of the leg occurs in fact under the knee. Consequently internal/external rotation assistance is not necessary either.

3.3.3 Weight

Unlike the devices that act on all the legs articulations (e.g. Ekso, ReWalk, HERCULES, REX,...) partial devices cannot carry their mass by themselves. Indeed, the device load does not have a direct contact with the ground and therefore the reaction forces have to pass through the human body. Note that this is the case even for ankle devices, at least during swing phases. Consequently, the mass of these devices have to be reasonable. Studies about the influence of added masses carried at different body locations have been conducted to estimate the maximal acceptable values for such wearable devices (Meuleman et al., 2013)(Browning et al., 2007)(Rossi et al., 2013). It appears that the proximal loads have a limited impact on gait parameters when compared with distal loads. It can reasonably be assumed that loads of 8% and 2.7% of body mass located respectively at the trunk and the ankle levels are the maximal values in order not to perturb the energetic and kinematic gait parameters (Meuleman et al., 2013). These values are valid for healthy adults and have not been tested with elderly people. However their order of magnitude is assumed to be fairly adequate for subjects receiving assistance.

3.3.4 Dynamics

The most dynamic activity that the typical users of assistive devices may want to perform is walking. It is very unlikely that a person being able to jump or run wearing a robotic orthosis would need such a device to walk normally. Standard gait trajectories of healthy adults (see 1.3.2) are therefore used to develop the dynamic requirements. By deriving the position we estimate that the joints velocities in the sagittal plane are typically around 150 $^\circ/s$ for the hip and 300 $^\circ/s$ for the knee. The second derivative enables to deduce the Root Mean Square (RMS) accelerations which are around 1500 $^\circ/s^2$ and 4000 $^\circ/s^2$. Even though the hip and the knee may have different trajectories in elderly it is assumed that these variations are reasonable. This assumption is further supported by the study presented in appendix B.

3.3.5 Torque and power

Assisting gait essentially means following the articulations trajectories and providing a certain percentage of the torque that is required to perform these movements. The torques at the different joints are presented in 1.3.3 (see Fig. 1.4). At the hip, the peak torque typically reaches

0.8 $N \cdot m/kg$ of body mass while the knee requires about 0.5 $N \cdot m/kg$. As the objective is not to mobilize the user's leg, a maximum of approximately 30 to 40% of assistance appears reasonable for studying assistive strategies.
Some gait related activities such as standing up from a chair require more torque than level walking. This action typically requires peak torques of about 1 $N \cdot m/kg$ of body mass at the hip and at the knee (Mak et al., 2003). Stair climbing/descending is also more demanding than walking, especially at the knee level where the peak torque can reach more than 1 $N \cdot m/kg$ of body mass (Riener et al., 2002).
The specifications in terms of power at the hip and knee are estimated from typical walking mechanical power curves recorded on elderly[4] (Eng and Winter, 1995). Here again as only a percentage of assistance is to be provided the peak values can be reduced to 30–40% of that of the reference[5]. The peak value at the hip is therefore around 0.4 W/kg of body mass. The assistance at the knee needs to provide approximately 0.15 W/kg of body mass when generating mechanical work and 0.4 W/kg when dissipating energy.

3.3.6 Transparency

The first requirement regarding the general behavior of an assistive device is that it must be able to appear mechanically transparent to the user. This control mode is commonly called zero impedance[6]. This means that the impact induced by the robotic orthosis (and in particular its actuation units) on the user should ideally be unperceivable. The gravitational, inertial and frictional effects should then specifically be compensated for, as they are usually the most important ones. This can be implemented with a full compensation. In that case a precise model[7] of the hardware is required. If a force sensor is available, a feedback loop can also be implemented to provide a null output force. Note that transparency is not only a matter of control; the available sensors and the mechanical design also play a major role in that context.
The transparent mode is essential as the target users are able to move and their movements should not be restrained especially in the case where no assistance is provided. However, perfect compensation appears idealistic and in practice moderate resisting torques are anyway acceptable. Similarly to what is presented in 3.3.3, we can assume here that the human beings are unaffected by reasonable perturbations when walking. In (Meuleman et al., 2013) it is

[4]Due to the redistribution of the articulation torques induced by aging, considering values recorded on young subjects would be inaccurate. However, the values are not dramatically different and these small differences can clearly be neglected when dimensioning the actuation units.

[5]The velocities are strictly the same but the torque is reduced according to the percentage of assistance.

[6]The mechanical impedance, in analogy to the electrical impedance, describes the force created by a structure (associated to voltage in the electrical domain) when subjected to a velocity (≡ current). In the analogy, a mass m corresponds to an inductance and therefore creates a force proportional to the acceleration (≡ derivative of the current) $F_{mass} = m \cdot \frac{dv}{dt}$, where v is the imposed velocity; a damper d corresponds to a resistance and consequently creates forces that are proportional to the velocity $F_{damper} = d \cdot v$; eventually a spring k corresponds to a capacitor and thus generates forces proportional to the position (≡ integral of the current) $F_{spring} = k \cdot \int v \, dt$.

[7]This model is either calculated from physical data or identified

Table 3.1 – Summary of the technical requirements for an assistive orthosis.

Articulation	RoM [°]	Max. velocity [°/s]	Max. RMS acceleration [°/s^2]	100 % assistance torque [Nm/kg]	100% assistance power [W/kg]	Maximal resisting torque [Nm]
Hip flexion extension	130	150	1500	0.8 when walking 1 when standing up	1 when generating power 0.4 when dissipating power	3.5
Knee flexion extension	140	300	4000	0.4 when walking 1 when standing up 1.4 when climbing or descending stairs	0.4 when generating power 1.3 when dissipating power	3.5

reported that a mass of 0.3 kg located at the ankle has a visible effect on the hip motion[8]. We therefore assume that an inertia corresponding to a point mass of 0.3 kg located at the foot already has an effect on gait. For an individual measuring 1.70 m, this corresponds to an inertia of approximately 0.25 $kg \cdot m^2$ (it is assumed here that this individual has legs of 85 cm). As the average RMS acceleration at the hip during normal walking is around 800 $°/s^2$, a resisting torque of 3.5 $N \cdot m$ is induced. As this resisting torque has a moderate effect on gait, we assume that the sum of all the undesired torques of an assistive device must be smaller than 3.5 $N \cdot m$. The hip and the knee being able to generate torques of the same order of magnitude, this value is used for both articulations.

3.3.7 Summary

Table 3.1 summarizes the main technical requirements for hip and knee assistive orthoses developed specifically for elderly. These requirements are different than that conventionally used for paralyzed patients. The presented designs (see chapter 5) are then based on these specifications which are adapted for the seniors.

[8]This result is partially in contradiction with what is described in 3.3.3 where much heavier masses have a impact on gait that is not considered significant. In fact, Meuleman and colleagues concluded that these changes induced by relatively limited masses are negligible as they do not exceed the intra subject variability.

4 Mechanical concepts

The prototypes that were developed as part of this thesis are all based on electric motors as this technology stays the most convenient way to actuate a wearable device[1]. Even though other types of actuation have been developed and successfully integrated into orthotic devices (e.g. pneumatic artificial muscles (Ferris et al., 2006), linear hydraulic actuators (Zoss et al., 2006)) the electric motors are still easier to integrate and control. Rigid mechanical structures are also used for similar pragmatic reasons. Here again alternative exists with the so-called soft exosuits but at the moment these lines are being written only preliminary results obtained with such devices have been published (Bae et al., 2015).
Important mechanical concepts related to the actuation and to the structure are recalled in the following sections. Theory about mechanical transmissions is concisely presented. As interaction between the user and the orthotic device is fundamental, the power transmission between the two has to be handled properly. Basic concepts of parallel kinematics are also introduced as orthotic devices work in parallel with the limb they assist. Therefore they cannot work as if they were isolated and closed kinematic loops must be considered.

4.1 Motor and transmission

The order of magnitude of the torque that is typically required for an orthotic device is of tens of $N \cdot m$. The velocity of the joints are around few hundreds of $^\circ/s$. The corresponding power can then easily be reached with electric motors of a decent size. However, such actuators rotate much faster and can only provide a small fraction of the required torque. A transmission is therefore mandatory.
A mechanical system with a motor, a transmission and an end-effector can be represented as a black box as presented on Fig. 4.1. The energy flow defines which end of the transmission

[1]Hydraulic and pneumatic actuation units are often more noisy than electric motors. Hydraulics is subject to leaks which cause cleanness problems and may be dangerous in case the liquid performs at high temperature. Pneumatic is also subject to loss. Even though they are less hazardous than that of hydraulic actuators, they tend to compromise the efficiency and therefore a compressor is often required to run continuously.

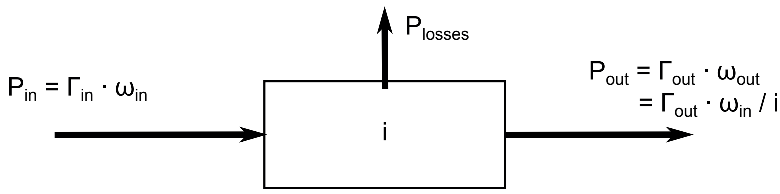


Figure 4.1 – Black box representing a transmission. The output velocity is equal to the input velocity rated by the transmission ratio. The power is transferred to the output with losses. The output torque is consequently affected by these losses.

is the input. When a motor is used to accelerate a load, it is the input of the system. It is the most common way of using an actuator. With an assistive device, the energy flows both ways as is the case with haptic interfaces. When the motor torque is opposed to the velocity of the load (or the action of the user), the motor has to be seen as the output of the system. Note that in that case it dissipates energy. In any case the input power P_{in} circulates to the output through the transmission. The transmission ratio i between the input velocity ω_{in} and the output velocity ω_{out} is given by:

$$i = \frac{\omega_{in}}{\omega_{out}} \tag{4.1}$$

In the idealistic case where no dissipative effects exist, the torque/force is rated with the inverse of the transmission ratio. Theoretically the power would then be the same at the input and at the output. Note that this is never the case in practice. With friction, only part of the torque is transferred to the output and some power P_{losses} is dissipated. Moreover, the friction depends on the load, the velocity and also possibly on the position, the orientation of the mechanism, the temperature, etc. In the case of non-back-drivable transmissions, the energy can be transmitted in one direction only (in the velocity reduction direction). The forces applied at the end-effector are then completely dissipated in the mechanism and not transferred to the motor side. It can be proven that such systems have an efficiency inferior to 50% in their normal direction of use (under the hypothesis that the dissipative power is proportional to the load) (Spinnler, 1980).

In the case of pure mobilization devices a non-back-drivable transmission is acceptable as the user is not expected to move by himself. It may even be useful to maintain a fixed position (e.g. in a squat position). In that case, no motor torque is required and the required electrical power is null. On the other hand, for assistance, a non-back-drivable transmission is not adapted as the user cannot act on the motor side. Bidirectional interaction is then impossible unless additional sensors are used to detect the user's motor intentions.

The friction on the motor side $\Gamma_{motor\_friction}$ is rated by the transmission ratio to be seen at the end-effector as:

$$\Gamma_{motor\_friction,end-effector} = \Gamma_{motor\_friction} \cdot i \tag{4.2}$$

The inertia is also affected by the transmission ratio. With the energy conservation law it can be demonstrated that the inertia of the motor I_{motor} is perceived at the end-effector as:

$$I_{motor,end-effector} = \frac{I_{motor}}{\eta} \cdot i^2 \tag{4.3}$$

with η being the efficiency of the transmission.
The transmission ratio has then to be defined carefully in order to ensure a proper bidirectional interaction. Friction also needs to be avoided at the most so as to optimize the efficiency of the transmission.

4.1.1 Reduction with fixed transmission ratio

As the required torque is usually around two orders of magnitude higher than the motor torque, a corresponding reduction ratio is required. With most methods (e.g. belts, gear,...) a ratio between two diameters is used to create a reduction. Several stages of reduction are then required to achieve an appropriate ratio. The last stage on the end-effector side needs to be robust enough in order not to be damaged. Note that in the case where more than one stage are used, the first one(s) do not need to bear as high constrains as the last one and should therefore be designed lighter.
Belt, chains, ropes, cables, and gears are the most commonly used technologies for transmitting a rotational movement.
Belts may typically be used in orthotic devices for the first stage of reduction. However, they are rarely used as the last one as the required torques would imply very bulky belts, either very wide to bear high tension or long and mounted on large pulleys.
The chains are heavier and noisier than the belts and require lubrication. For those reasons they are generally not used in wearable devices. However, a last stage of reduction would be possible with this technology.
In some cases, ropes or cables can advantageously replace the belts or the chains. Conventionally the cables and the ropes are attached at their extremities, which implies that the range of movement is limited. Note that in the case of orthotic devices, the movements need to be bounded in any case (see 3.3).
Gears can be used in many different configurations and then offer a wide range of solutions. A high ratio may be obtained with epicyclic gear trains. They are therefore frequently used by motor manufacturers. With the Harmonic Drive technology (Harmonic Drive AG, *Germany*), an even higher transmission ratio can be achieved in the same volume.
In the case of the Harmonic Drive, one reduction stage is usually enough. When it is not the case, the different technologies may be combined to achieve the best tradeoff limiting the weight, the inertia and the frictional effects while ensuring a suitable robustness.

4.1.2 Reduction with variable transmission ratio

The requirements in terms of torque and maximal velocity vary depending on the activity that the user may want to perform. We remind here that for instance the torque that is required at the knees and hips when standing up from a sitting position is larger than when walking on a flat terrain (see 3.3). The sit-to-stand transition is typically a movement that may require assistance, even with people who are able to walk unassisted. Moreover this torque is maximal when the flexion of the knees and the hips are around 70°. A reduction ratio that varies with the position may therefore be an advantageous feature.

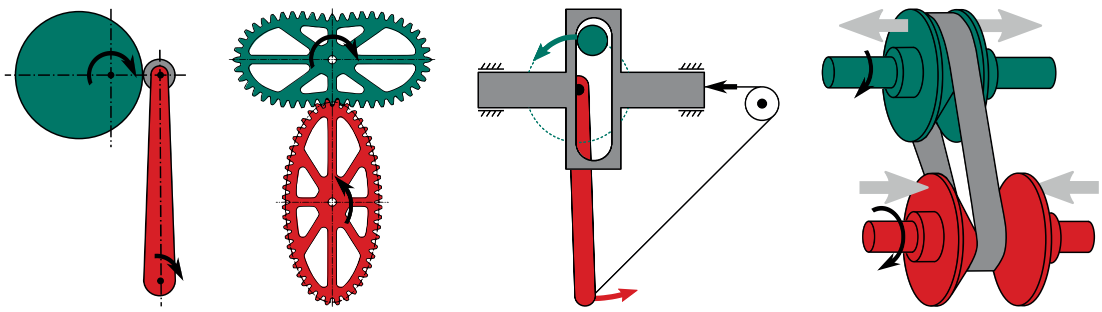
Another type of transmission with a variable ratio associates the reduction to the velocity. Even though the joints torques are not directly correlated with the velocities at the different articulations during walking (see section 1.3), assistance is rationally needed with people walking slowly. If a person is able to walk fast, logically no assistance is necessary.

Variable transmission ratios can be obtained with cam mechanisms (see Fig. 4.2(a)). Special gears can similarly be used to create the same effect (see Fig. 4.2(b)). In principle any transformation of the movement will create a transmission ratio that may possibly vary with position. Linkages can be used to shape the transmission ratio that is desired. Multiple movement transformations may be required to obtain the desired profile (i.e. transmission ratio as a function of the position). The motion may even be reversed (i.e. opposed to the arbitrarily decided forward direction) if the transmission ratio becomes negative. With devices intended to assist gait, this feature can be particularly helpful as a continuous rotational movement may be transformed into cyclic movements. In (Ryder and Sup, 2013) a Scotch-Yoke mechanism (see. Fig. 4.2(c)) is used to create this effect. The advantage is that the actuator only needs to vary slightly its velocity to create the desired output walking pattern. A large amount of energy that would be used to accelerate and decelerate the mechanism inertia is thus spared. The evident drawback of this type of mechanism is that the movement is imposed in-between fixed boundaries. Note that this undesired effect was limited in Ryder and Sup's design by using a spring placed in-between the actuation and the part attached to the leg.

The variation of the transmission ratio may also be controlled actively. Ideally the variation should be smooth so as to avoid discontinuities in the transmitted torque. Continuously Variable Transmission (CVT) can then be used to optimally adapt the transmission ratio. Fig. 4.2(d) shows a typical way of implementing such a transmission. The reduction may then depend on the position but similarly on many other parameters according to the velocity and torque requirements at a specific point in time.

4.1.3 Series elastic and variable stiffness actuators

Conventionally the interface between an actuator and a load is built as stiff as possible. This tradition comes from the industrial robots that need precision, stability and high bandwidth position control. However, having a compliant interface may have some advantages including tolerance to shocks, better capabilities in terms of force control, capacity to store elastic energy and lower reflected inertia (Pratt and Williamson, 1995). These benefits are of major



(a) (b) (c) (d)

Figure 4.2 – Variable transmission ratio mechanisms. (a) Cam mechanism. The output movement and the transmission ratio depend on the position of the cam. (b) Gears with variable transmission ratio. The principle is similar to the cam's. Almost any ratio is possible as long as the distance between the centers of rotation remains constant. (c) Scotch-Yoke mechanism as implemented in (Ryder and Sup, 2013). The rotational movement of the input is transformed to a linear movement which is then used create a cyclic movement at the output. (d) Transmission with cones and belt. The transmission ratio can be modified by adjusting the positions of the two pairs of cones.

interest when it comes to orthotic devices in particular for assistance. Series Elastic Actuators (SEA) have therefore been introduced for such purposes in many applications (Blaya and Herr, 2004)(Giovacchini et al., 2013). A spring in series with the actuation unit (i.e. after the motor and the transmission as presented on Fig. 4.3) enables to treat the output force control problem as a position control problem, which is much easier to perform especially when a reduction with a high ratio is used. Moreover, no force sensor is required, which consequently limits the overall cost.

An improvement of the SEA is possible by making the stiffness of the series elasticity adjustable. The Variable Stiffness Actuators (VSA) enable to regulate the output stiffness by means of a second actuator. A possible way to implement a VSA is by using non-linear springs (see Fig. 4.3). When the two springs are tensioned, an effect similar to the co-contraction of biological muscles appears and the stiffness increases. Note that this effect would not appear with linear springs. The tension difference between the two springs makes it possible to vary the set position. Another way to obtain this effect is with lever arms acting on linear springs as presented in (Jafari et al., 2011) or (Groothuis et al., 2012).

4.1.4 Transmission with clutch

An important capability that an assistive device should have is the transparency (see 3.3). Indeed, assisting consists mainly in following the user's movements and make them easier to perform. While various control strategies may exist in order to define how this assistance is provided to the user, transparency appears to be an ideal starting point.

Following the movements of the joints cyclic trajectories requires a significant amount of energy. The rotational acceleration of the motor is usually much higher than the joint's because

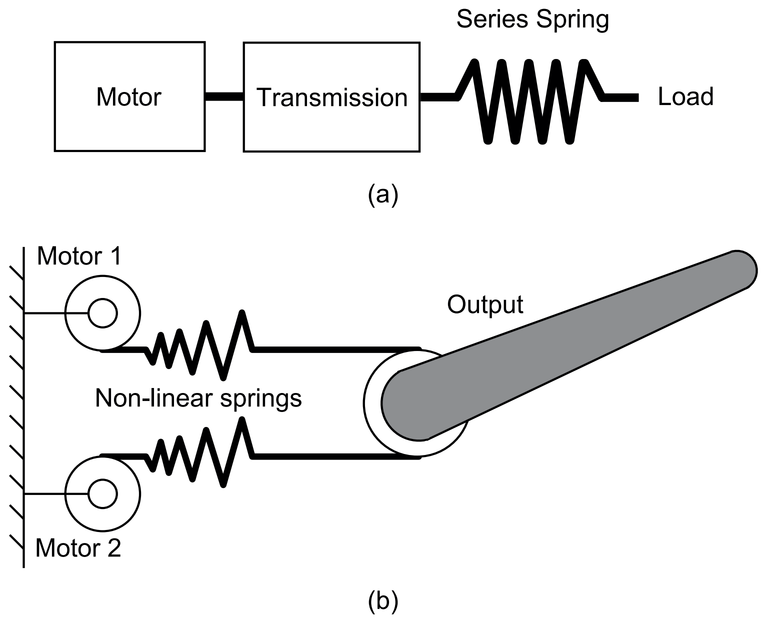


Figure 4.3 – Schematic representations of series elastic and variable stiffness actuators. (a) A spring is placed in series with an actuator and a transmission. (b) Two non-linear springs may be used to adjust the stiffness perceived at the output. In that case two actuators are required. The set point is defined by moving the two actuators in the same direction while the stiffness is adjusted by a differential action. Note that for the sake of readability, a simplification concerning the representation of non-linear springs is introduced in this figure. The actual implementation of a non-linear spring is in fact slightly more complex.

of the transmission. This effect is even greater at fast cadences. A trivial way to tackle this problem is to decouple the motor and the leg by using a clutch (see Fig. 4.4(a)). Usually a clutch transfers torque from a motor to an end-effector. In fact, the torque is transferred from the end that acts as a velocity source. The other end is seen as a mechanical impedance. A clutch can then provide torque only in one direction which depends on the velocity difference between the input and the output. Consequently if the motor end is slower than the leg's, only a dissipative torque can be provided. On the other end, if it is faster it can only generate positive torques. If the two ends are at the same velocity the torque is transferred as in a standard coupling. In this case the efficiency is optimal as no loss due to friction exists. When there is sliding, the efficiency η_{clutch} decreases and becomes:

$$\eta_{clutch} = \frac{\omega_{out}}{\omega_{in}} \tag{4.4}$$

with ω_{in} being the velocity of the fastest end and ω_{out} the velocity of the other end.
In a mechanism with a motor, a transmission and a clutch (actively controlled) only the inertia on the end-effector of the clutch is perceived by the user (unless the clutch is completely locked). Indeed, the torque transferred by the clutch depends on its control and is therefore not proportional to the acceleration (unless voluntarily defined this way). Another consequence of the uncoupling is that as long as the motor end rotates faster than the end-effector side, torque can be generated. The motor and the transmission therefore act as a flywheel and the instantaneous available peak power is higher than the motor's (without overcharge required). Note that similarly, in the case of a slow motor end, a great dissipative torque can be generated. With non-back-drivable transmissions this torque can be as high as is allowed by the mechanical resistance of the various components.

4.1.5 Transmission with double clutch and inversion mechanism

With a single clutch, torque in one direction only can be supplied. In order to be able to instantly generate torque in one direction or the other two clutches need to be used. An inversion mechanism is then required so as to be able to reverse the motion for one of the two clutches (see Fig. 4.4(b)). One constraint remains: the end-effector velocity needs to be smaller than the motor rated velocity; otherwise only dissipative torque can be provided. This concept was first demonstrated in (Fauteux et al., 2010). In this design, differentials and electrorheological brakes are used to implement the clutches. One of the differentials is also used to reverse the motion (see Fig. 4.4(c)).

4.1.6 Transmission with a combination of spring, clutch and inversion mechanism

Elastic energy storage and clutching principle can be combined in order to limit the waste of energy during the negative work phases. In (Haeufle et al., 2012), the authors propose a mechanism to improve the actuators capabilities in the context of prosthetic and legged devices. In their design an electric motor can be coupled with a spring that acts in parallel with

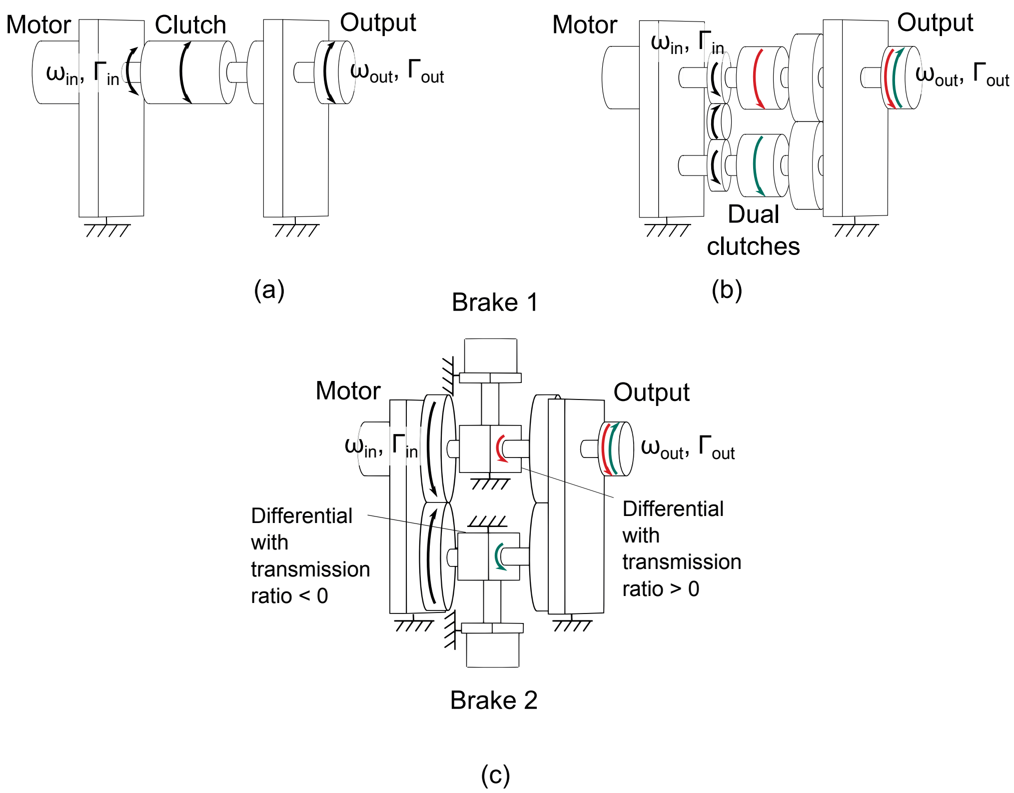


Figure 4.4 – Clutch mechanisms. (a) Standard clutch. The output torque is adjusted by the clutch. Its direction depends on the velocity difference between the input and the output. (b) Double clutch with inversion mechanism. The output torque can be in both direction provided that the input velocity is higher than the output's. (c) Double differential clutch. It works similarly to the double clutch but the torque is controlled through brakes.

it. It is then a Parallel Elastic Actuators (PEA) controlled by a clutch. Note that the reflexion on PEA only is not developed here as no demonstration has been done that they may offer significant benefits for orthotic devices. However, the action of the parallel spring may help the actuator in particular for hopping activities. In conventional PEA implementations, the effect of the parallel stiffness cannot be adjusted during operation. With the Clutched Parallel Elastic Actuator (C-PEA) design, the spring can be uncoupled when not needed.
More recently, the Bi-directional Clutched Parallel Elastic Actuator (BIC-PEA) was proposed to control the loading and the unloading of the parallel spring (Plooij et al., 2015). With the BIC-PEA, the energy can be stored and also returned in both direction thanks to a differential and two locking mechanisms. Moreover, this energy can be released when it is needed, which means that the mechanism can be locked. An additional benefit is that the torque does not depend on the position as would be the case with a standard PEA. Indeed, the energy storage can start at any position.
In (Stramigioli et al., 2008) a concept is presented that aims at storing at the most the energy that is normally dissipated with conventional designs. The idea is to use a non-back-drivable actuator in series with a spring. Stramigioli proposes to use a brake on the actuator output shaft in order to guarantee its non-back-drivability. In-between the spring and the end-effector an Infinite Variable Transmission (IVT) is used for controlling the output torque[2]. Consequently, no clutch is required anymore as the uncoupling can be performed by the IVT itself. Indeed a transmission ratio that reduces the end-effector movements to zero on the motor side can be seen as a decoupling. The actuator is used to load the spring when this is not passively done by the end-effector during the negative work phases. The IVT is in charge of converting the torque that the spring naturally provides into the desired output torque (positive or negative). Fig. 4.5(a) presents this configuration. While the concept appears promising, no practical realization has been manufactured yet.
An alternative to the IVT is possible with a double clutch and an inversion mechanism (see Fig. 4.5(b)). With such a mechanism only two transmission ratios (one positive and one negative) are possible. The maximal positive power that can be generated is then limited by the torque available at the spring level (possibly rated by a stage of reduction before the end-effector). On the other hand, negative work can be as high as needed as the excess energy can be dissipated by using the two clutches simultaneously (one of them locked and the other partially open). This concept is further developed in section 5.2 where it is integrated in a robotic knee orthosis.

4.2 Kinematics for orthotic devices

An orthotic device acts in parallel with the limb it assists. Consequently one or more kinematic loops are created, which usually affects the overall mobility. Furthermore, the humans articulations are complex joints which, most of the time, cannot be considered as conventional mechanical joints. First because they are made of biological structures that may have irregular

[2]An IVT is a CVT whose range of transmission ratio includes positive, null and negative values.

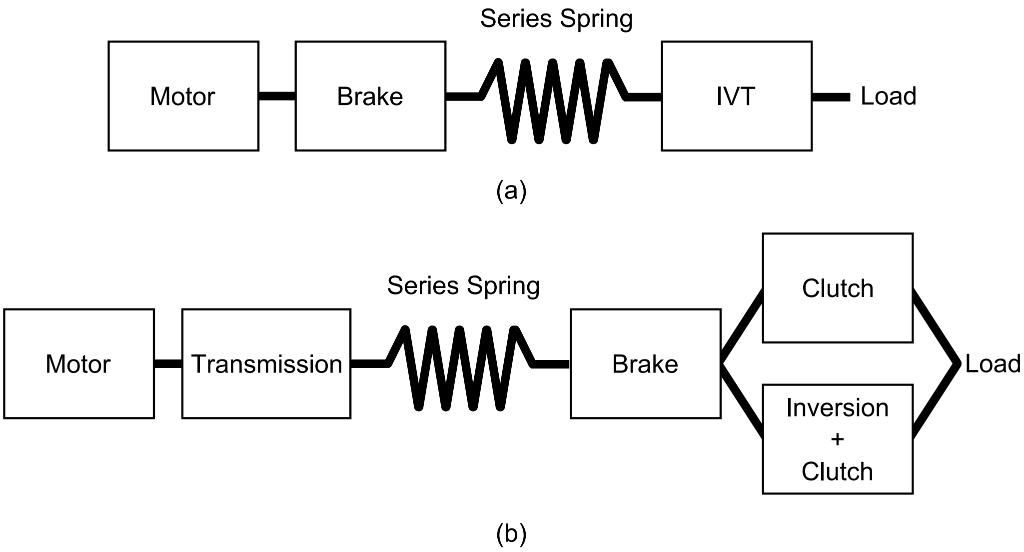


Figure 4.5 – Schematic representations of transmissions with spring and inversion mechanism. In theses mechanisms, a spring is loaded by a motor and is used to store energy. (a) The end of the spring that is on the motor side can be locked by a brake. The other end is employed as a torque source. An IVT is then used to adjust the output torque to the desired value. (b) The brake can be removed if a non-back-drivable transmission is used. A back-drivable transmission can as well be used but a torque needs to be provided by the motor to maintain the desired position. The IVT can be replaced with a double clutch and an inversion mechanism. The torque that the spring generates can then be transmitted in both directions but the maximal output torque when providing energy is the torque of the spring. The torque that can be provided when dissipating energy is higher as both clutches may be used simultaneously.

shapes and second because they are enclosed in soft tissues (i.e. muscles, tendons, ligaments, fat and skin). For those reasons, the kinematics of the orthotic structure is crucial. We remind here that assistive devices are ideally not constraining any movements (see section 3.3 and subsection 2.1.5).

When a loop exists in a kinematic chain additional constraints appear which may reduce the overall mobility of the system. In order to keep the initial mobility of a joint when adding elements creating a loop, two methods can be considered. First, a precise adjustment can be done between two kinematic chains working in parallel in order to align their respective constraints. If the two chains constrain the same movements they theoretically do not limit the movements of the other. Such designs are called over-constrained and they are usually not recommended in mechanical design as the proper functioning of the system depends a lot on the manufacturing and on the stability of the constraint alignments. Second, additional Degrees of Freedom (DoFs) can be added in the design so as to compensate for the misalignment of the constraints. Both of these methods will be developed hereafter.

The prototypes and concepts that were developed within this thesis focus mainly on the hip and, to a lesser extent, on the knee. Our attention turns therefore mainly to these two joints. However, most of the reasoning developed in this subsection can be applied to other articulations.

Unlike most of the biological articulations, the hip joint can be considered as a simple mechanical joint, i.e. a ball and socket joint (see 1.3). The major difficulty to address with this joint is that it has three DoFs with relatively large Range of Motion (RoM). Moreover, the center of rotation (i.e. the head of the femur) is inside the body which makes it difficult to precisely localize.

The knee has less DoFs than the hip. It is often considered that only the movement in the sagittal plane exists even though other rotations can be observed (see 1.3). Their RoM is quite limited though. The particularity of the knee is that its flexion/extension movement is not only a rotation around a fixed axis. Indeed, the knee movement is better approximated by considering that the tibia rolls on the femur, which makes the axis of rotation move (see 1.3).

When designing an orthotic device, it is also important to consider the overall system composed by the user and the orthosis. Indeed, a device that assists more than one articulation cannot usually be considered as two devices each assisting one joint. Depending on the fixations on the user the constraints may vary a lot.

4.2.1 Adaptation to the biological movements

Hip joint

The hip can move around the head of the femur whose shape is almost spherical. Three pure rotations are therefore considered around this point. A mechanism that mimics these rotations therefore needs three rotational DoFs. In order to obtain an over-constrained system hip plus orthosis that is yet able to move, precise alignments is required between the axes of

rotation of the mechanism and the hip. For now let us consider the neutral position being the standing position (without any flexion/extension, adduction/abduction or internal/external rotation).
The axes of rotation in the sagittal and the frontal planes can be aligned relatively easily with an external mechanism. Indeed the horizontal axes passing through the head of the femur intersect the body close to the articulation. The device can then be placed close to the joint it assists. For the third axis it is more complex as it is aligned on one side with the leg and on the other with the upper part or the body. A Remote Center of Rotation (RCR) mechanism is then strongly recommended.
The straightforward approach to obtain three rotational DoFs is to have three rotational joints in series, each of these joints being initially aligned with the x, y, z axes defined on Fig. 4.6 (a). Six different permutations therefore exist depending on which of the three rotations comes first, second and third. The second rotation in the chain is the most critical. Indeed, while the first and the third rotation do not change the orthogonality of the three axes, the second does. The first rotation applies a rotation to the two other DoF (see Fig. 4.6 (b)). The third axis does not influence the two first DoFs (see Fig. 4.6 (d)). The second rotation influences only the third DoF and therefore makes it move relatively to the first one. Fig. 4.6(c) illustrates this effect. A singularity is then reached when a $\pm 90°$ rotation of the second joint aligns the third axis with the first one. A wise choice for the second DoF is therefore a rotation that has a limited RoM in order to avoid any singularity.
The alignment of the mechanical joints with flexion/extension, adduction/abduction and internal/external rotation is not necessary. However, the alignment makes it easier to control each rotation independently. Note that this is the case only if the mechanical joints remain aligned with their corresponding human DoFs. The third joint is therefore always aligned. The second joint is aligned only when the third joint is in its neutral position. The first joint logically requires that the second and the third joints are in their neutral positions. The implication for under-actuated mechanisms is then quite important. If only one of the human DoFs requires assistance, the assisted joint should ideally be the last in the kinematic chain. In practice this rule does not have to be strictly respected for two reasons. First, the effects of the other DoFs are usually negligible as their RoM are limited for most Activity of Daily Living (ADL) (sitting represents an exception as the flexion angle is usually quite important in this posture). Second, the assistance may still be beneficial if it does not follow strictly the planned trajectory. We remind here that the definitions of the of the principal movements are themselves ambiguous. For example, the flexion/extension is defined as the rotation in the sagittal plane. When combine with other rotations this definition becomes unclear as the sagittal planes of the trunk and the leg are different. If the trunk sagittal plane is considered, the flexion of the hip would therefore be aligned with a joint that is always parallel to x (see Fig. 4.6 (a)) and thus first in the kinematic chain.

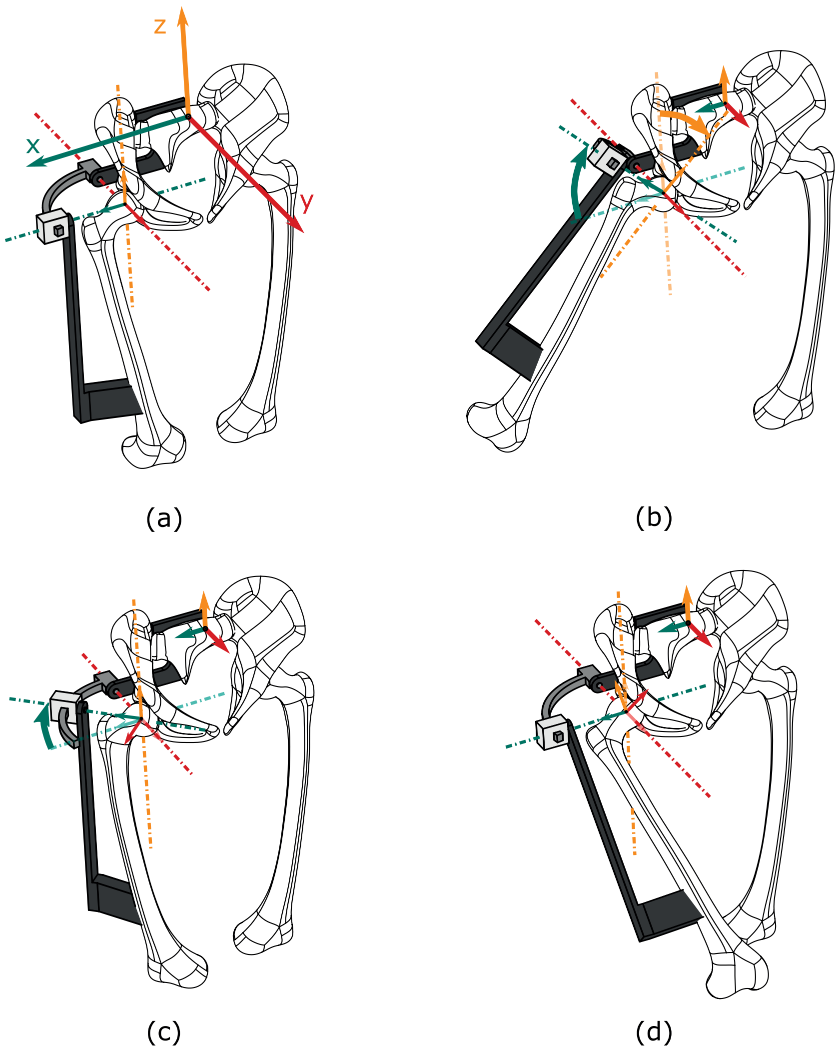


Figure 4.6 – Kinematics of an over-constrained system hip and 3 DoFs orthosis. (a) In the neutral position the three axes of rotation of the device are ideally orthogonal and aligned with the hip's axes. (b) When rotating the first axis of rotation the two other axes rotate and they all stay orthogonal to each other. (c) When the second axis is moving, the third one rotates with it while the first does not. The orthogonality is therefore not preserved. This may create a singularity when the first and the third axes are merged. (d) When a rotation about the third axis of rotation occurs the axes remain orthogonal as it does not influence any other axis.

Knee joint

The knee joint is mainly composed of two bones: the femur and the tibia. The end of the femur may be considered spheroidal while the end of the tibia is almost planar. While this model is relatively simple, it appears valid in most cases even though more complex models exist (Sancisi and Parenti-Castelli, 2010)(Walker et al., 1985). The movement is therefore a rotation of a sphere on a plane. The out-of-plane movements are constrained by tendons, ligaments and muscles. This constraint is usually considered rigid. Consequently, the movement of one point on the end of the femur has only one DoF which is defined by a cycloid. Then the trajectory cannot be approximated by a pivot as a non-negligible translatory movement exists. Several methods exist to mimic the trajectory of the knee joint. First, a straightforward implementation would be to reproduce the shapes of the two bones (or at least their projection in the sagittal plane). In order to guarantee a rolling without sliding, a rack and pinion may be used. In standard orthopedic devices two gears are used for ease of implementation. Indeed, the distance between the center of the rolling gear needs to be fixed relatively to the surface of the reference element (rack or second gear). With gears this can be done very easily by linking the two gears centers together whereas a linear guiding would be required with rack and pinion.

Another commonly used method consists in using a crossed four-bar linkage as presented in (Bertomeu et al., 2007) or (Tucker et al., 2013). By selecting correctly the lengths of the different linkages, a relatively good fitting may be obtained. Tucker et al. typically obtained an error on the center of rotation position of maximum 1.05 *mm*. This error would be critical in a conventional mechanism as rigid components would need to deform in order to align the two kinematic chains trajectory. This would create high elastic forces, lock the mechanism or even damage it. In orthotic devices the two kinematic chains are always linked together through soft tissues whose stiffness is much lower than conventional mechanical parts'. The elastic force due to the deformation induced by a 1 *mm* misalignment is then negligible.

Interface on deformable tissues

The compliance at the interface with the body due to soft tissues may be helpful when misalignment appears. Even though the stiffness is anisotropic, a displacement of few *mm* is normally possible in any direction without creating high forces. Similarly few degrees of deformation are permitted around all axes as they will only generate negligible torques. The position of the mechanical structure may then move a little relatively to the the biological joint.

The relative displacement of the orthosis is not always advantageous. With the deformation of the biological tissues that appears when moving or when the muscles contract, the mechanism may also lose its alignment with the articulation. These displacements may be quite large (typically around a couple of *cm*). The realignment of the two kinematic loops may then generate high forces and torques.

4.2.2 Kinematics with additional DoFs

When no parasitic force is permitted on the body tissues, they may be considered rigid. The misalignment compensation must then be handled differently. In any coupling of rigid bodies the Chebychev-Grübler-Kutzbach criterion applies. The mobility formula states:

$$M = 6 \cdot (N - 1 - j) + \sum_{i=1}^{j} f_i \tag{4.5}$$

where, M is the mobility of the system, N is the number of links, j is the number of joints and each joint i possesses a freedom f_i. Note that here we consider that one of the links is a fixed reference. If only one articulation is assisted (e.g. the hip or the knee), the number of part is three: the two body parts plus the orthosis. The number of link is also equal to three as both body parts are attached together and to the device. As the initial mobility of the joint should be equal to the mobility of the system, the device must have 6 DoFs. Note that any additional kinematic chain that adds a loop locks 6 DoFs. A mechanism that assist all three articulations of the leg may have only 6 DoFs if attached only at the pelvis and the foot. On the other hand, if it is attached to a third point on the body, a second kinematic loop is created and 12 DoFs are required.
Depending on the kinematic design, some DoFs can be considered as the main movements while the others are used to compensate for the parasitic movements. Such mechanisms usually look like the over-constrained mechanisms as they have their main DoFs aligned with the human articulations. The only difference is that they have additional joints to compensate for the trajectory errors. With these designs the additional kinematic chain (i.e. the mechanism) can be close to the body as their movements are very similar. Wearable devices usually do not have kinematics that differ completely from the articulation they assist for that reason. However, with additional DoFs the articulations with complex kinematics (e.g. the knee) can be assisted with simple joint(s). The misalignment joints compensate then for the trajectory inaccuracies.

5 Solutions for articulation assistance

In section 5.1 two hip orthoses are presented. Both designs were developed considering the concepts presented in chapter 4. The first one is actuated by a ball-screw mechanism which provides a reduction with a variable transmission ratio while the second uses a double-clutch and an inversion mechanism.
In section 5.2 a knee orthosis concept is detailed. Even though this device was not manufactured, it is worth presenting as its transmission has interesting features. Moreover, the knee being different than the hip or the ankle in terms of assistance needs, the hardware should take into account these differences and be designed accordingly.

5.1 Assistive hip orthoses

In this section two prototypes are presented. The first one is based on a variable transmission ratio mechanism (see 4.1.2). The transmission is optimized to be maximal when a high torque is required (i.e. when the flexion angle is around 70°). In the walking RoM it is smaller in order to enable relatively fast movements. The kinematics is based on additional DoFs and is therefore not over-constrained.
The second prototype uses a transmission with a double differential and brakes mechanism, which behaves similarly to a double clutch and inversion mechanism (see 4.1.5). The kinematics of this device is also based on additional DoFs but has a different configuration than the first orthosis.
The technical capabilities of each prototypes are contextualized for diverse activities in order to better compare them. Objective characteristics such as the RoM, the transparency, the maximal torque that they can provide or the Root Mean Square (RMS) torque during cyclic trajectories are compared to point out which device is better adapted for specific situations.

5.1.1 Hip Ball-Screw Orthosis (HiBSO)

Part of this subsection is adapted from (Olivier et al., 2013b) and (Olivier et al., 2014b).

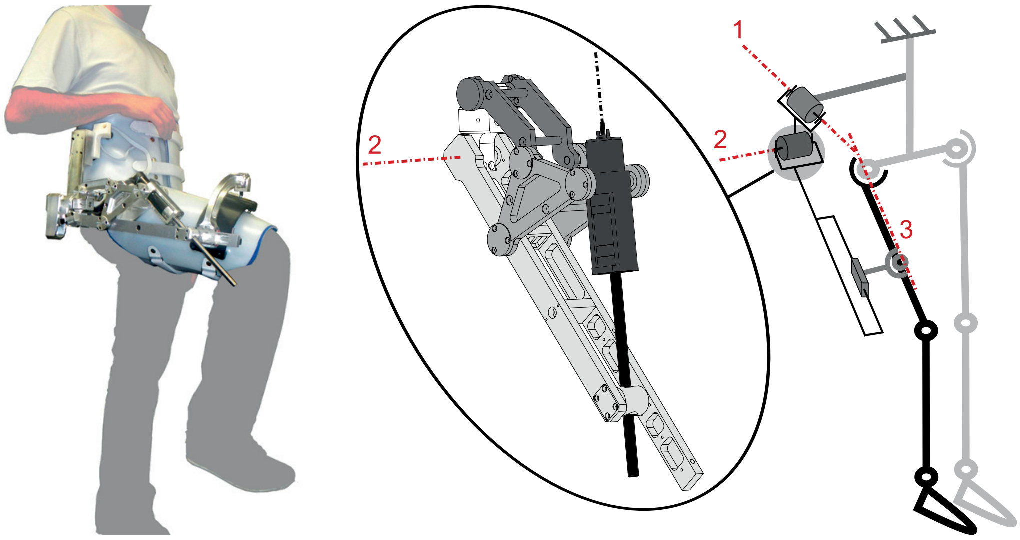


Figure 5.1 – Hip Ball-Screw Orthosis (HiBSO). Left: actual prototype worn by a healthy user. Right: Kinematics of the system user and device. Middle: The actuated DoF is second in the device kinematic chain and uses an amplification mechanism based on a ball-screw.

Actuation

In assistive orthoses the energy flows both ways between the human and the device. A back-drivable mechanism is therefore required as the user may need to drive the actuator (see 4.1). Back-drivability largely depends on efficiency, which also needs to be high in order not to dissipate energy unnecessarily. Linkages represent a very efficient way to transmit motion as friction can be limited at the most. Moreover, a variation of the transmission ratio depending on the position is possible with this method.
For the first stage of transmission, a spindle drive (ball-screw) from Maxon was selected for its excellent efficiency (around 94%) and for the high output force it can provide (1 *kN*). The pitch of the ball-screw is 2 *mm*. The combination of the spindle drive with linkages is also of great interest. Indeed, the limited travel of the screw may restrict the overall RoM. Using linkages as employed in excavators (see Fig. 5.2) enables both the variation of the transmission ratio and a large RoM.

Movement conversion On Fig. 5.3 the final design is presented with its different parameters. These parameters were used to tune the movement conversion between the motor and the end-effector which approximately corresponds to the leg flexion angle (see 5.1.1 kinematics, below). This was performed so as to achieve a large transmission ratio (~ *1*: *300*) around 70°. In this area, high torque is required for the sit-to-stand transitions. In the walking RoM, the ratio varies between ~ *1*: *120* and ~ *1*: *250*. Moreover, according to the the commonly used walking trajectories (Winter, 1991), the maximal velocity and acceleration occur right after

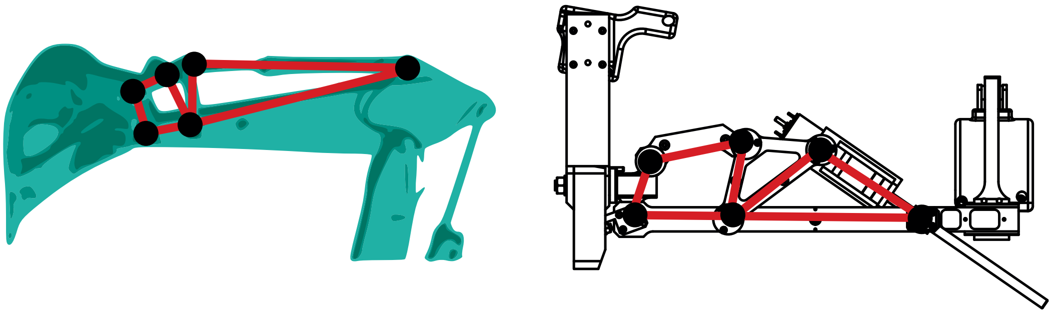


Figure 5.2 – Excavator and amplification mechanism comparison. The limited RoM of the linear actuation unit is used to produce a relatively large rotational output movement.

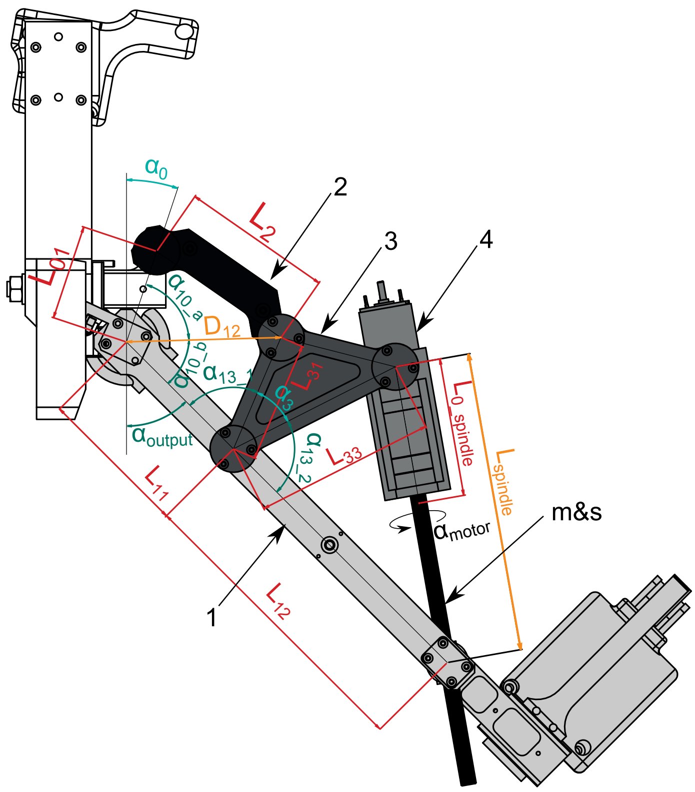


Figure 5.3 – Computer-Aided Design (CAD) side view of the orthosis with important parameters used in the equations. The moving parts attached together have the same shade. Fixed lengths are displayed in dark red, variable lengths in orange, fixed angles in cyan, and variable angles in dark turquoise.

the full extension phase of gait (when flexion is still limited). A relatively small transmission ratio is therefore preferable as it limits the inertia perceived from the end-effector (see 5.1.1 dynamic capabilities, below).
The forward kinematic model is expressed hereafter. First, the angle of the motor α_{motor} makes the length $L_{spindle}$ vary:

$$L_{spindle}(\alpha_{motor}) = L_{0\_spindle} + \frac{p}{2 \cdot \pi} \cdot \alpha_{motor} \tag{5.1}$$

where p is the pitch of the ball-screw. $L_{0\_spindle}$ is defined on Fig. 5.3. The other parameters are calculated with the following equations:

$$\alpha_{13\_2}(L_{spindle}) = acos\left(\frac{L_{spindle}^2 - L_{33}^2 - L_{12}^2}{-2 \cdot L_{33} \cdot L_{12}}\right) \tag{5.2}$$

$$\alpha_{13\_1}(L_{spindle}) = \pi - \alpha_3 - \alpha_{13\_2} \tag{5.3}$$

$$D_{12}(L_{spindle}) = \sqrt{L_{11}^2 + L_{31}^2 - 2 \cdot L_{11} \cdot L_{31} \cdot cos(\alpha_{13\_1})} \tag{5.4}$$

$$\alpha_{10\_a}(L_{spindle}) = acos\left(\frac{L_2^2 - L_{01}^2 - D_{12}^2}{-2 \cdot L_{01} \cdot D_{12}}\right) \tag{5.5}$$

$$\alpha_{10\_b}(L_{spindle}) = acos\left(\frac{L_{31}^2 - L_{11}^2 - D_{12}^2}{-2 \cdot L_{11} \cdot D_{12}}\right) \tag{5.6}$$

Eventually, the end-effector angle is given by:

$$\alpha_{output}(L_{spindle}) = \pi - \alpha_0 - \alpha_{10\_a} - \alpha_{10\_b} \tag{5.7}$$

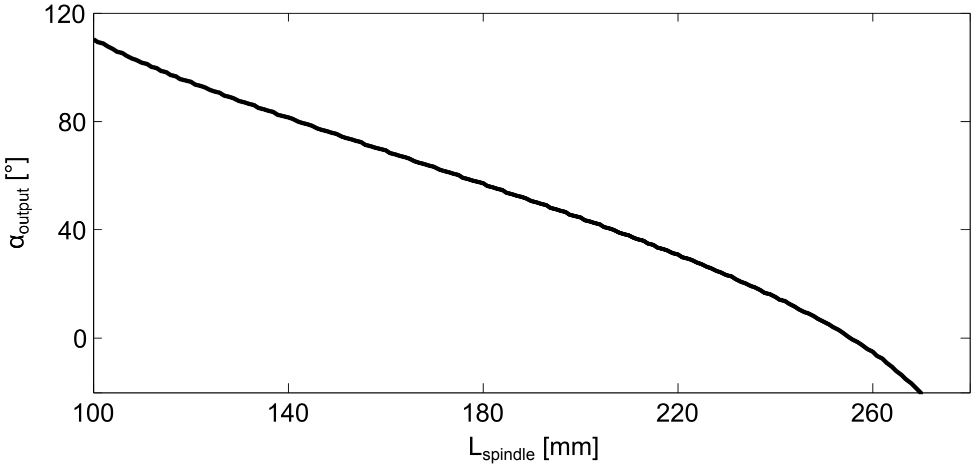
The transmission ratio is then calculated directly from the inverse of the Jacobian which expresses the relation between the angular velocities of the actuator and the end-effector:

$$\dot{\alpha}_{output} = J \cdot \dot{\alpha}_{motor} \tag{5.8}$$

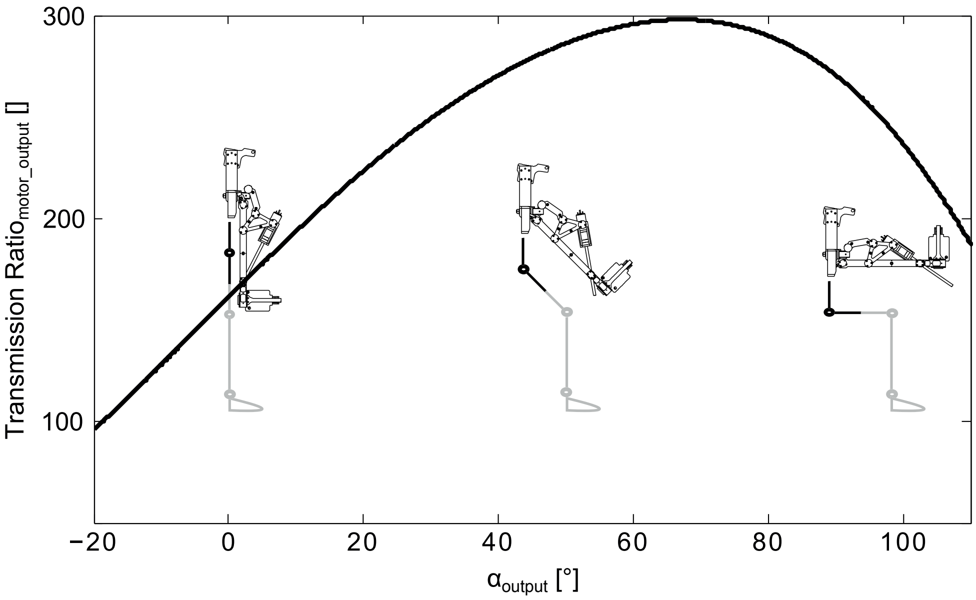
The transmission ratio TR is therefore:

$$TR = J^{-1} = \frac{\dot{\alpha}_{motor}}{\dot{\alpha}_{output}} = \frac{1}{\frac{\partial \alpha_{output}}{\partial \alpha_{motor}}} \tag{5.9}$$

Fig. 5.4(a) shows the relation between the spindle length $L_{spindle}$ and the end-effector angle α_{output}. The transmission ratio TR as a function of the latter is shown on Fig. 5.4(b).



(a)



(b)

Figure 5.4 – Characteristics of the transmission. (a) Output angle as function of the spindle length. (b) Transmission ratio between the motor and the output as function of the output angle. The different parts of the design (see Fig. 5.3) were dimensioned by simulation so as to shape this function according to the needs.

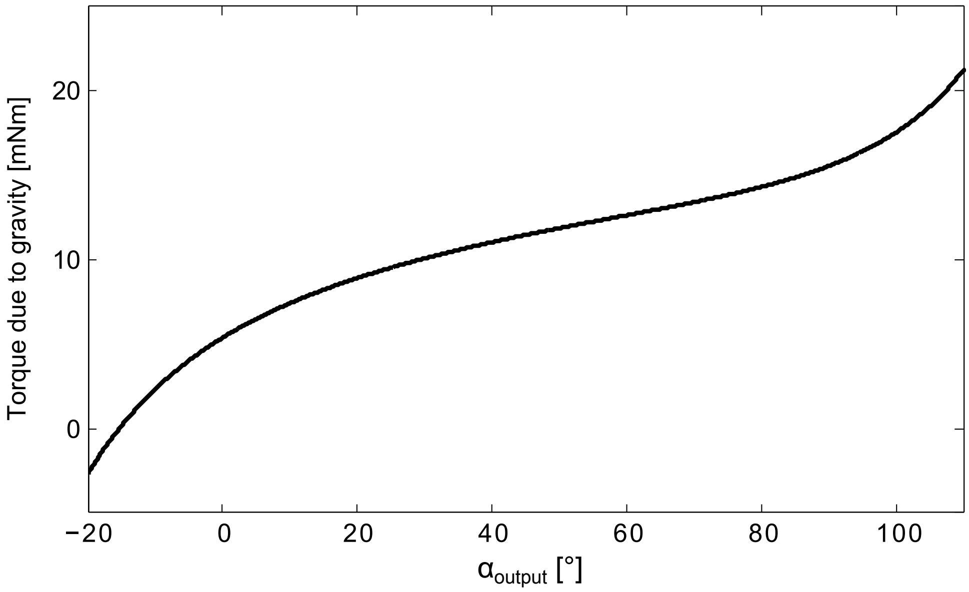


Figure 5.5 – Torque due to gravity. The different moving parts have an influence on the gravitational torque. The variable transmission ratio then rates this torque that is perceived on the motor side.

Dynamic capabilities The level of assistance that the orthosis can provide varies as a function of the power that it uses for moving its own structure. If a lot of power is used to perform rapid trajectories, only little torque can be provided to the user for assistance. Several forces need to be counteracted so as to follow a desired trajectory. First, friction must be compensated for. Then, the mass of the different parts imply that gravity will have an effect on the overall structure. The mass also impacts on the accelerations as the different movements are influenced by the rated inertias of the different parts composing the orthosis.

The friction in the mechanism is very low and is mainly due to the friction of the motor itself. The output rated torque due to friction is typically around 1 to 2 $N \cdot m$, which represents about 6 $mN \cdot m$ at the motor side.

The gravitational torque is neither null (as it would be the case with a balanced system) nor sinusoidal which would be the case with a simple pendulum. By solving the static equations for each moving part, a precise estimation of the gravitational effects is calculated. In order to be able to use this estimation to actively compensate for gravitational forces, the torque required at the motor level to get a statically stable system is also calculated. This torque is presented on Fig. 5.5.

Unlike mechanisms with a fixed transmission ratio, the torque required to make our orthosis move depends on position and velocity in addition to acceleration. This is intuitively explained by the fact that even at constant output velocity, some parts in the mechanism accelerate or decelerate. The changing geometry also implies changes in inertia. The different shades of grey on Fig. 5.3 show the parts that are moving together. Sub-assembly 1 (light grey on Fig. 5.3) has a constant effect on the inertia perceived at the output level. Therefore, the torque required to make it move is expressed by:

$$M_{output\_1} = I_{output\_1} \cdot \ddot{\alpha}_{output} \tag{5.10}$$

where I_{output_1} is the inertia of the assembly 1 seen from the axis of rotation. The other moving parts induce torques M_i which are perceived at the output level as follows:

$$M_{output\_i} = J_{i\_output} \cdot M_i + I_{output\_i} \cdot \ddot{\alpha}_{output} \tag{5.11}$$

where i is the number of the considered assembly (1 to 4 or the motor and screw $m\&s$), J_{i_output} is the jacobian linking the angular velocities (or the torques if 100% efficiency is assumed) of the output and of the assembly i, and I_{output_i} is the inertia of assembly i about the output axis of rotation. The torques M_i are expressed as:

$$M_i = I_i \cdot \ddot{\alpha}_i = I_i \cdot \frac{d(\dot{\alpha}_i)}{dt} = I_i \cdot \frac{d(J_{i\_output} \cdot \dot{\alpha}_{output})}{dt} \tag{5.12}$$

$$= I_i \cdot \left(J_{i\_output} \cdot \ddot{\alpha}_{output} + \frac{\partial (J_{i\_output})}{\partial \alpha_{output}} \cdot (\dot{\alpha}_{output})^2 \right) \tag{5.13}$$

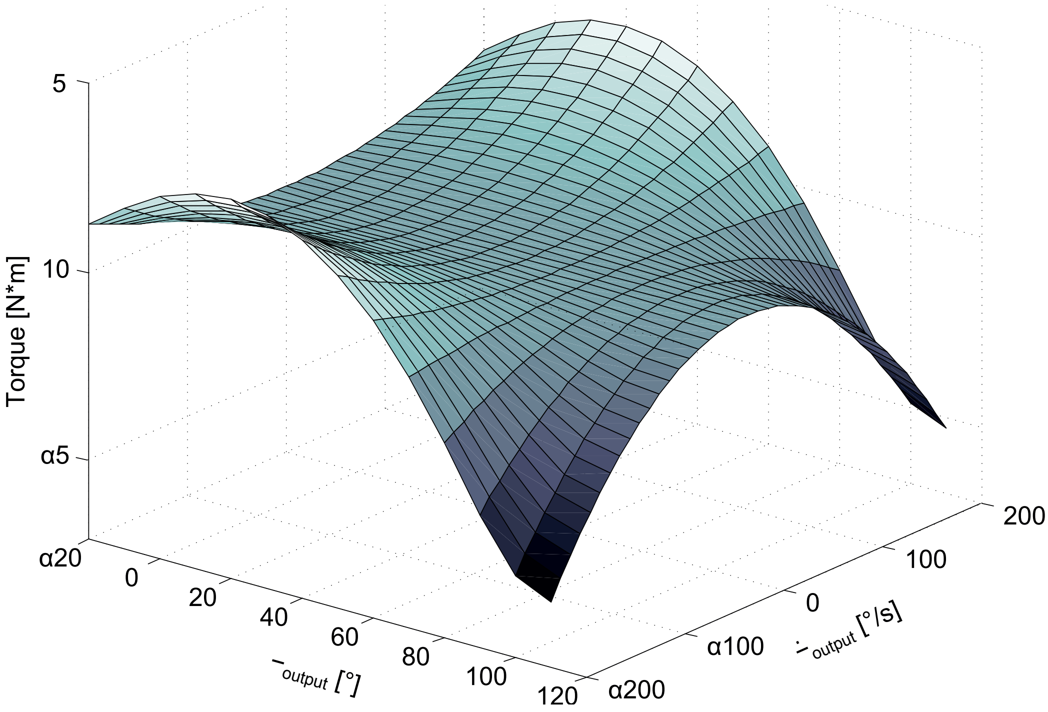
where I_i is the inertia of assembly i about its own axis of rotation.
The most important contribution to inertia comes from the motor and the screw $m\&s$. It is given by:

$$M_{output\_m\&s} = J \cdot I_{m\&s} \cdot \left(J \cdot \ddot{\alpha}_{output} + \frac{\partial J}{\partial \alpha_{output}} \cdot (\dot{\alpha}_{output})^2 \right) \tag{5.14}$$

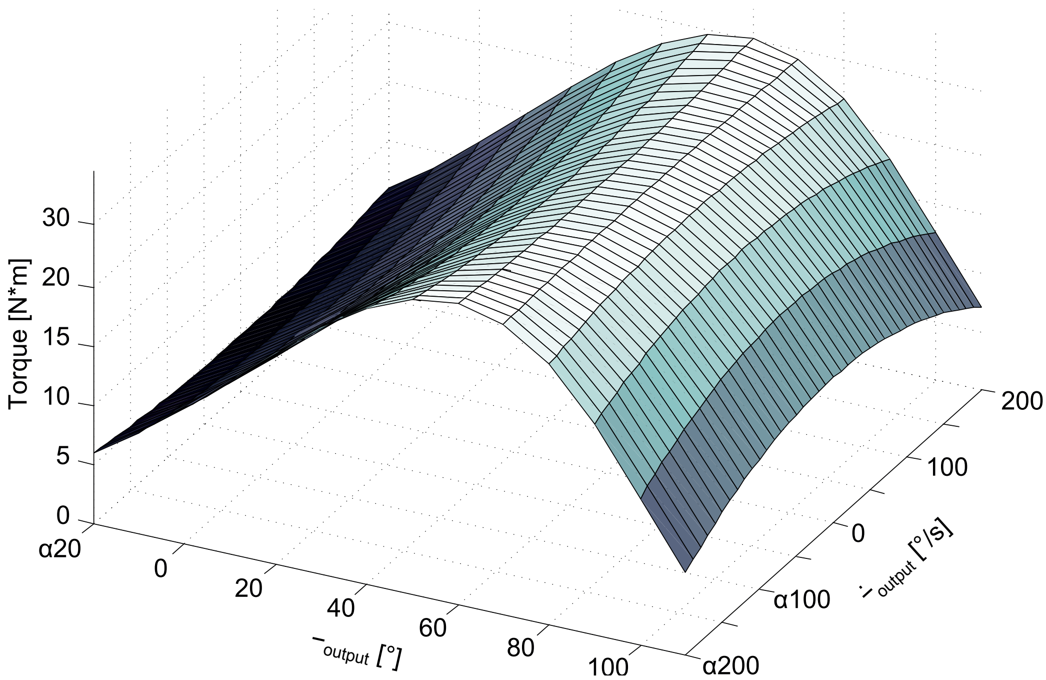
Eventually, the torque due to all the different orthosis parts is:

$$M_{output} = \sum_{i=1}^{4} M_{output\_i} + M_{output\_m\&s} \tag{5.15}$$

Because of the non-linear transmission ratio, this torque depends not only on the acceleration but also on the position and the velocity. Fig. 5.6 presents the torque due to inertia that is required to move the orthosis at two different accelerations. Fig. 5.6 (a) is at a constant velocity (acceleration is equal to zero) and Fig. 5.6 (b) is at an acceleration of 2000 $^\circ/s^2$.
In order to test the actual dynamic capabilities of the HiBSO, a typical walking trajectory was used as a reference to track. Different walking cadences (corresponding to the number of steps per minute) were used in order to measure how it influences the torque required by the motors to follow the trajectory. Fig. 5.7 (a) shows the trajectory as a function of the stride cycle. With the implemented control, the orthosis follows precisely this input at the different tested cadences. The motor torque used to follow this trajectory at a cadence corresponding to 100 *steps*/*min* is shown on Fig. 5.7 (b). The RMS torque spent by the motor to accelerate the mechanism is 43.5 *mNm*, which corresponds to about 50% of its nominal torque. Under these conditions, the mechanism would be able to provide 30% of assistance for a 70 *kg* person. The tests performed at higher cadences demonstrated that no extra torque can be provided by the motor above a walking frequency of 1 *Hz* (corresponding to 120 *steps*/*min*) (see Fig. 5.7 (c)).



(a)



(b)

Figure 5.6 – Torque due to inertia. (a) When the orthosis moves at a constant velocity. (b) When the orthosis has an acceleration of 2000 $^\circ/s^2$.

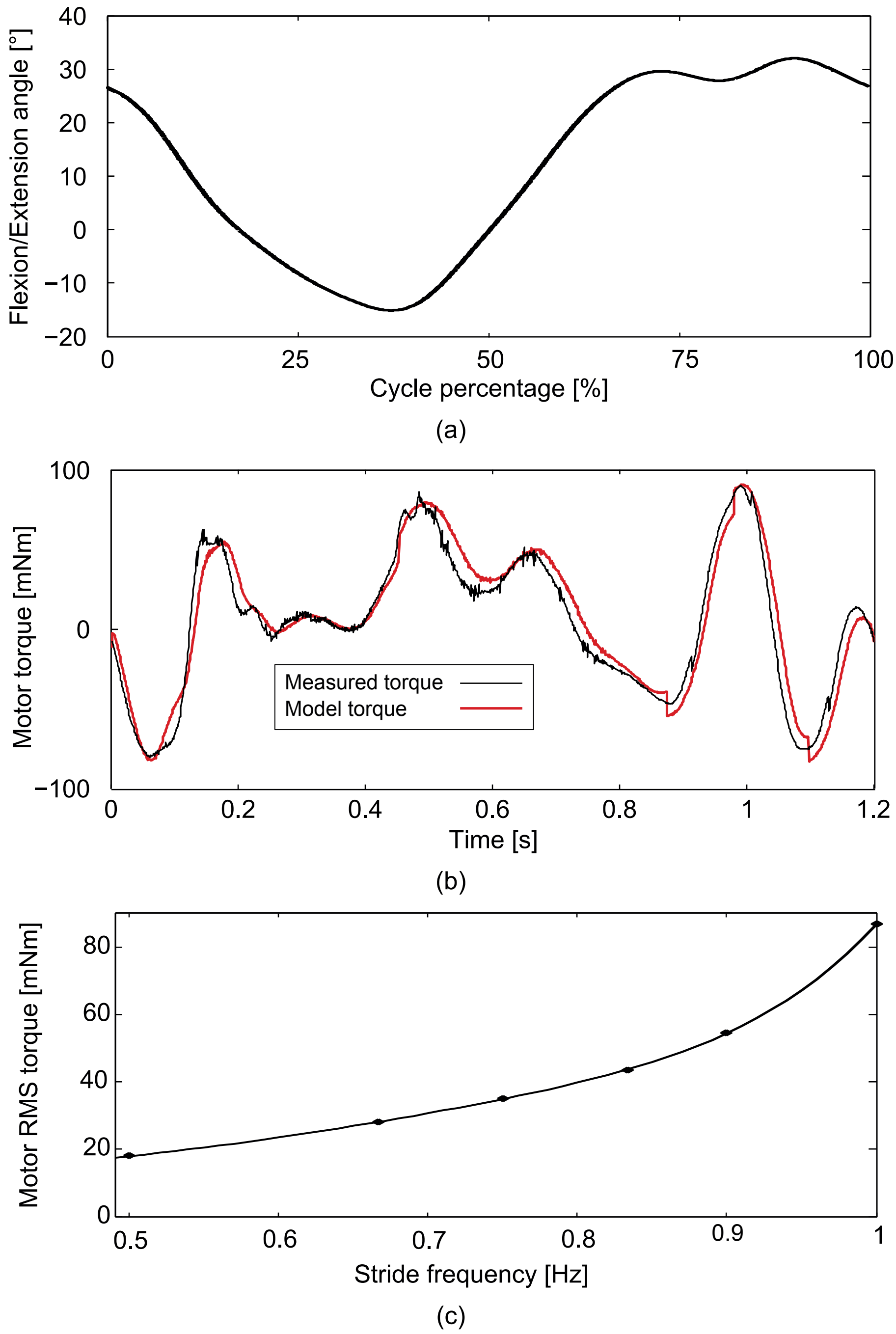


Figure 5.7 – Dynamic capabilities of the device on typical walking trajectories. (a) Walking trajectory used for testing the mechanism capabilities. (b) The torque during this cycle at a frequency of 0.83 *Hz* was recorded and compared to our model. The torque RMS value is 43.5 *mNm* which corresponds to about 50 % of the nominal motor torque. (c) RMS value of the motor torque during the trajectory at different frequencies. The maximum frequency is around 1 *Hz*. In that case no assistance can be provided.

Kinematics

The kinematics of the HiBSO is based on additional DoFs so as to avoid over-constraints (see 4.2.2). In its first version, 6 DoFs are used. Two joints are located at the interface with the pelvis. They approximate the movement in the sagittal and in the frontal planes. The other four DoFs are at the thigh level. They are obtained with one spherical and one prismatic joints. The actuated DoF being relatively bulky, it cannot be placed first in the kinematic chain. Contrary to the recommendation stated in 4.2, the actuated joint which corresponds to flexion/extension is placed second in the kinematic chain. The first joint corresponds initially to adduction/abduction. This is not advised as flexion has a large RoM and it may then move the structure to a singular position where the joints that should correspond to the first and the third human DoF become parallel. In this position, the mechanical joints that enables adduction/abduction is parallel with the leg and therefore the movement in the transverse plane is impossible[1]. On the other hand, the rotation about the axis aligned with the thigh is enabled by two mechanical joints which means that an internal DoF is created. To prevent this undesired effect, a cam mechanism is used so as to lock progressively the first DoF as a function of the second joint. One of the four DoFs at the thigh corresponds approximately to the internal/external rotation[2]. The other three DoFs are used to compensate for the misalignment of the main movements.
The mechanism is based on additional DoFs and therefore a precise alignment of the mechanical and the biological articulations is not necessary. The consequence of the non-alignment of the joints is that the match between the the main movements of the hip (flexion/extension, adduction/abduction, internal/external rotation) and the movement of the mechanism is not exact. This means that the joint that is supposed to correspond to adduction/abduction may have an influence on flexion/extension and on internal/external rotation. The same remark applies on the two other main joints.

Forward kinematics In this paragraph the first two joints of the mechanism are analyzed in details in order to express the actual effect of the device joints on the hip movements. To do so, the forward kinematics model linking the two first DoFs of the mechanism with the flexion and the adduction is derived. Fig. 5.8 presents the model used to approximate the HiBSO and its wearer. The different offsets respect the order of magnitude of the real device when worn by a user. The forward kinematics is calculated by localizing the intersection point of the human leg with the device. It is assumed that the fixation point on the thigh can move on a sphere whose center is the hip (head of the femur). The kinematic chain of the device is straightforward until the prismatic joint. The position of the latter cannot be assumed *a priori*. The position of the leg is therefore calculated by solving the equations describing a sphere

[1] It is reminded here that the reference planes do not move. The rotation about the axis aligned with the femur is then still possible. This rotation is about the antero-posterior axis when the flexion angle is 90°. This situation clearly demonstrates the limitation of the conventional terminology which uses projection on fixed planes.

[2] Evidently this applies only when the flexion/extension and adduction/abduction angles are null.

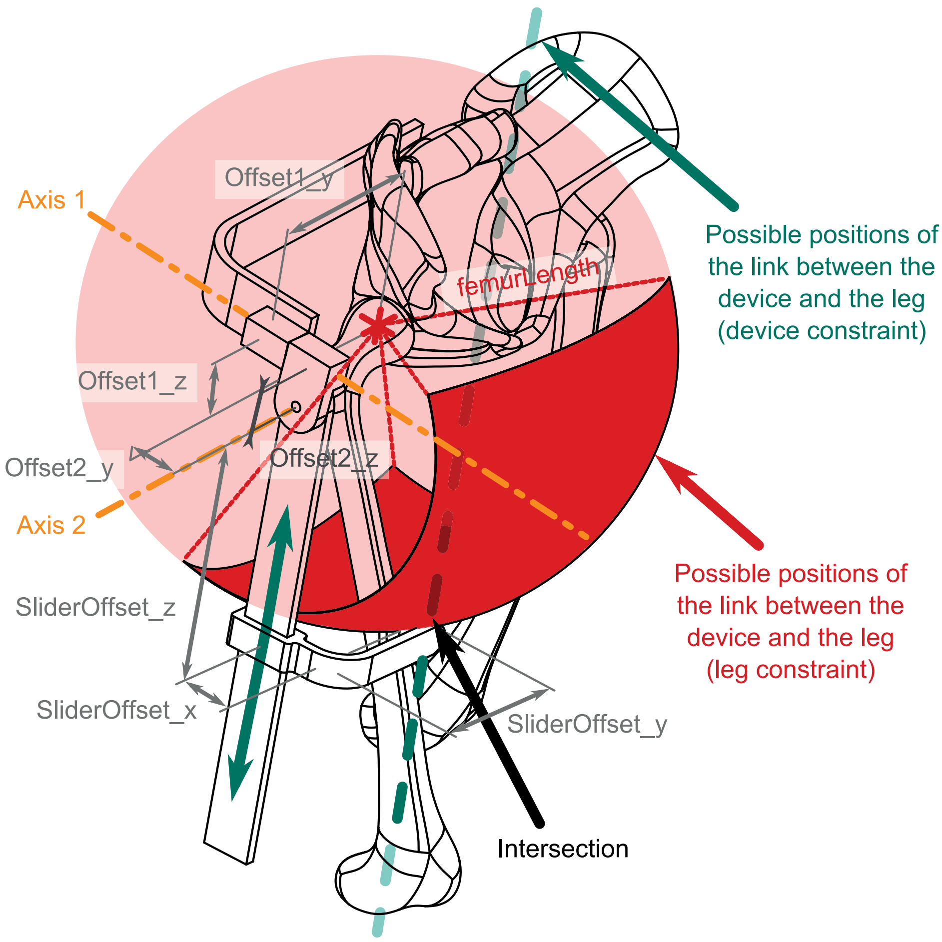


Figure 5.8 – HiBSO and wearer simplified model. The attachment point on the user's leg can move on the red sphere as its center of rotation is attached to the reference frame (i.e. the pelvis). The RoM of the hip limits the reachable area (dark red area). The end-effector of the device is defined by the position of the slider (turquoise dashed line) which also has a limited RoM (dark turquoise). When imposing the first two orthosis DoFs (orange axes), only one solution (flexion/extension and adduction/abduction) satisfies both conditions imposed by the turquoise line and the red area which fixes the position of the leg (the internal/external rotation of the leg is free).

(movement of the leg) and a line (possible movement of the device fixation at the thigh). If the line intersect the sphere and is not tangent to it, two intersection points exist. Only the solution that represents a possible physiological movement is then considered. This is done easily by considering the standard RoM of the hip (see Fig. 5.8). In case the ambiguity remains the RoM of the prismatic joint can also be used.
The first axis in the device kinematic chain has a fixed position which is defined relatively to the position of the head of the femur $\overrightarrow{hip}$ by *offset1_y* and *offset1_z*. A point on this axis is then defined by:

$$\overrightarrow{Axis_1} = \overrightarrow{hip} + \begin{pmatrix} offset1\_x \\ offset1\_y \\ offset1\_z \end{pmatrix} \tag{5.16}$$

The second axis moves with the rotation about the first axis. Its position depends on *offset2_x*, *offset2_z* and the rotation matrix $\mathbf{R_1}$ of the first axis which is defined by:

$$\mathbf{R_1} = \begin{pmatrix} 1 & 0 & 0 \\ 0 & cos(\theta_1) & -sin(\theta_1) \\ 0 & sin(\theta_1) & cos(\theta_1) \end{pmatrix} \tag{5.17}$$

where θ_1 is the angle of the first joint. A point on the second axis can then be calculated using the homogeneous matrix $\mathbf{H_1}$:

$$\mathbf{H_1} = \begin{pmatrix} 1 & 0 & 0 & \\ 0 & 1 & 0 & \overrightarrow{Axis_1} \\ 0 & 0 & 1 & \\ 0 & 0 & 0 & 1 \end{pmatrix} * \begin{pmatrix} & & & 0 \\ & \mathbf{R_1} & & 0 \\ & & & 0 \\ 0 & 0 & 0 & 1 \end{pmatrix} * \begin{pmatrix} 1 & 0 & 0 & \\ 0 & 1 & 0 & -\overrightarrow{Axis_1} \\ 0 & 0 & 1 & \\ 0 & 0 & 0 & 1 \end{pmatrix} \tag{5.18}$$

The initial position of a point on the second axis $\overrightarrow{Axis_{init\_2}}$ is defined by:

$$\overrightarrow{Axis_{init\_2}} = \overrightarrow{hip} + \begin{pmatrix} offset2\_x \\ offset2\_y \\ offset2\_z \end{pmatrix} \tag{5.19}$$

This point then moves with the first axis as:

$$\begin{pmatrix} \overrightarrow{Axis_2} \\ 1 \end{pmatrix} = \mathbf{H_1} * \begin{pmatrix} \overrightarrow{Axis_{init\_2}} \\ 1 \end{pmatrix} \tag{5.20}$$

The rotation matrix of the second axis is defined by:

$$\mathbf{R_2} = \begin{pmatrix} cos(\theta_2) & 0 & sin(\theta_2) \\ 0 & 1 & 0 \\ -sin(\theta_2) & 0 & cos(\theta_2) \end{pmatrix} \tag{5.21}$$

The corresponding homogeneous matrix $\mathbf{H_2}$ is then given by:

$$\mathbf{H_2} = \begin{pmatrix} 1 & 0 & 0 & \\ 0 & 1 & 0 & \overrightarrow{Axis_2} \\ 0 & 0 & 1 & \\ 0 & 0 & 0 & 1 \end{pmatrix} * \begin{pmatrix} & & & 0 \\ & \mathbf{R_2} & & 0 \\ & & & 0 \\ 0 & 0 & 0 & 1 \end{pmatrix} * \begin{pmatrix} 1 & 0 & 0 & \\ 0 & 1 & 0 & -\overrightarrow{Axis_2} \\ 0 & 0 & 1 & \\ 0 & 0 & 0 & 1 \end{pmatrix} \tag{5.22}$$

Eventually, the initial position of the prismatic joint is defined by:

$$\overrightarrow{Slider_{init}} = \overrightarrow{Axis_{init\_2}} + \begin{pmatrix} SliderOffset\_x \\ SliderOffset\_y \\ SliderOffset\_z \end{pmatrix} \tag{5.23}$$

Note that the parameter *SliderOffset_z* is not a constant as it varies with the position of the slider (i.e. the prismatic joint). The position of the link with the thigh $\overrightarrow{Intersection}$ is then given by:

$$\begin{pmatrix} \overrightarrow{Intersection} \\ 1 \end{pmatrix} = \mathbf{H_1} * \mathbf{H_2} * \begin{pmatrix} \overrightarrow{Slider_{init}} \\ 1 \end{pmatrix} \tag{5.24}$$

As the parameter *SliderOffset_z* cannot be known *a priori*, two arbitrary values are selected so as to define the line of all the possible points which can be reached with the device (see Fig. 5.8). This line is given by an initial guess $\overrightarrow{Slider_{initial\ guess}}$ and a vector $\overrightarrow{SliderVector}$ which is oriented according to the difference between the points calculated with the two arbitrary values of *SliderOffset_z*. The intersection point is then given by:

$$\overrightarrow{Intersection} = \overrightarrow{Slider_{initial\ guess}} + \overrightarrow{SliderVector} \tag{5.25}$$

The norm of the vector $\left\|\overrightarrow{SliderVector}\right\|$ and the elements of the vector $\overrightarrow{Intersection}$ are unknown. The other kinematic chain is the femur rotating around the hip. The point on the thigh that is attached to the device is therefore constrained on a sphere whose center is the head of the femur. The intersection point $\overrightarrow{Intersection}$ must therefore satisfy the following equation:

$$\begin{aligned} femurLength^2 = {} & (Intersection_x - hip_x)^2 + \\ & (Intersection_y - hip_y)^2 + \\ & (Intersection_z - hip_z)^2 \end{aligned} \tag{5.26}$$

where the indexes x, y and z correspond respectively to the first, the second and the third element of the vectors $\overrightarrow{Intersection}$ and $\overrightarrow{hip}$.
The four unknown $Intersection_x$, $Intersection_y$, $Intersection_z$ and $\left\|\overrightarrow{SliderVector}\right\|$ can then be solved using the equations 5.25 and 5.26. An analytical solution can then easily be found using any solver.
The thigh angles in the sagittal and the frontal planes are calculated from the difference between $\overrightarrow{Intersection}$ and $\overrightarrow{hip}$. The flexion/extension and the adduction/abduction angles are

then respectively given by:

$$\alpha_{flexion/extension} = tg^{-1}\left(\frac{Intersection_x - hip_x}{hip_z - Intersection_z}\right) \tag{5.27}$$

$$\alpha_{adduction/abduction} = tg^{-1}\left(\frac{hip_y - Intersection_y}{hip_z - Intersection_z}\right) \tag{5.28}$$

Fig. 5.9 (a) and (b) presents the flexion/extension and the adduction/abduction angles as a function of θ_1 and θ_2. As desired, the flexion/extension of the hip is very well correlated with θ_2 while θ_1 has almost no influence on it[3]. We remind here that this characteristics is very important when the device is under-actuated. The adduction/abduction is linked with θ_1 and is not dramatically influenced by θ_2 when the latter is smaller than 70°. When the flexion angle increases the adduction angle rises dramatically. This effect is mainly due to the fact that the definition of the adduction angle implies a projection on the frontal plane. When the flexion approaches 90° the vertical component decreases and the adduction angle becomes very sensitive to changes in the lateral direction. Fig. 5.10 shows this effect when the flexion angle is close to 90° (θ_2 is fixed at 100°). In this area one can notice that the leg movement imposed by the device is similar to a combination of flexion and internal/external rotation.
Fig. 5.10 also shows that the first axis of the device is unnecessary in this area (i.e. when the flexion angle is large) as it does not enable any movement that would not be possible without it. Even worse, this redundancy creates an internal DoF which may be disturbing for the user.

Cam mechanism With large flexion angles, the thigh tends to become parallel to the first axis of the device. Therefore a rotation about the axis that is perpendicular both to the leg and to the second axis becomes impossible. One of the three DoFs of the hip is then locked in this area. The second consequence is that the mechanism gains one DoF relatively to the user's body. This parasitic movement is not desired as the user's movements may be affected (e.g. by the moving masses).
A cam mechanism was thus designed so as to progressively lock the first device joint when the flexion increases. As the flexion is strongly correlated with the angle of the second orthosis joint (see Fig. 5.9) the latter can then be used to control the cam. The cam is composed by two surfaces corresponding to the extreme positions (see Fig. 5.11(a)). It is fixed to the reference frame (i.e. before the joints in the device kinematic chain). The cam follower is fixed after the second joints so that its position is influenced by the two rotations of the mechanism. The design of the cam was realized by using the homogeneous matrices $\mathbf{H_1}$ and $\mathbf{H_2}$ and several points on the cam follower (red part on Fig. 5.11(a)) to define the adequate surfaces.
The actual RoM of the first joint as a function of the second joint is shown on Fig. 5.11(b). When the flexion is smaller than 30° the maximal RoM is available and the first axis can rotate between –10° and 30°. From there, it gradually decreases until 60° in flexion. Above this flexion angle, the RoM is restricted to ±3°.

[3]This was tested with different initial parameters of misalignment corresponding to plausible values and very similar results were obtained.

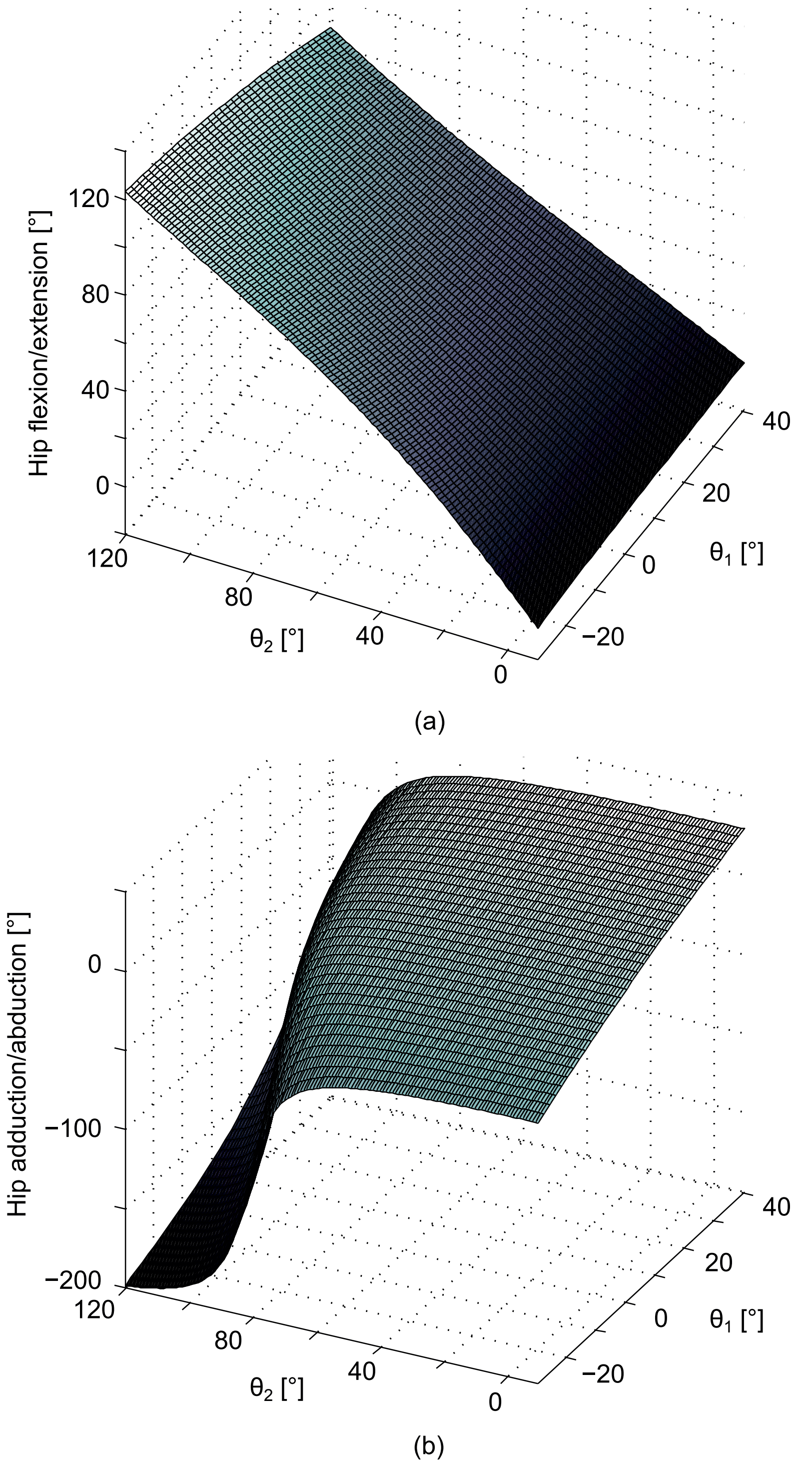


Figure 5.9 – Forward kinematics of the system HiBSO–human hip. (a) The flexion/extension of the leg is highly correlated with the second axis angle and not influenced dramatically by the action of the first joint. (b) The adduction/abduction is well correlated with the first rotation of the mechanism as long as the second joint angle is not too large. When the flexion angle is close to 90° the adduction/abduction becomes very sensitive to the device movements. This effect is mainly due to the ambiguous definition of adduction angle.

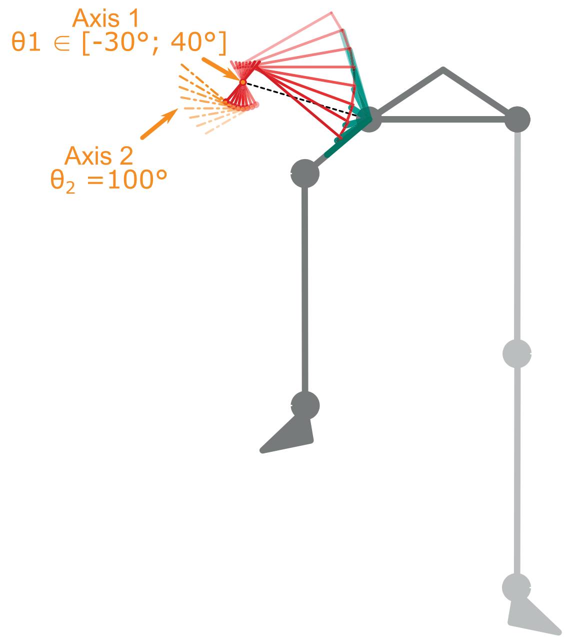


Figure 5.10 – Influence of a large flexion on the effect of the first joint on adduction/abduction.

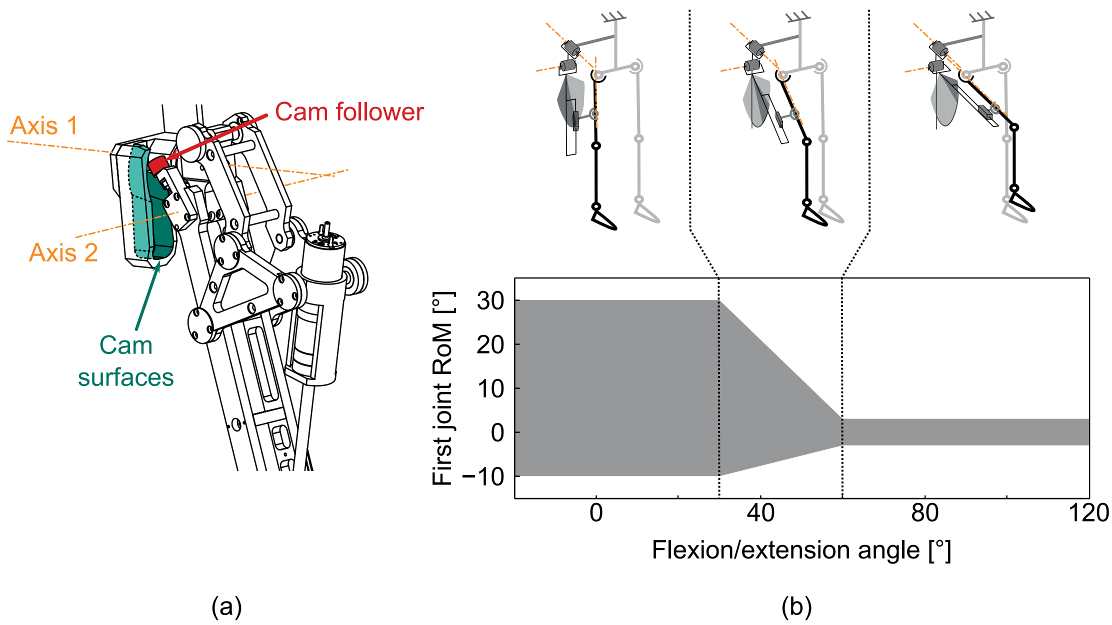


Figure 5.11 – HiBSO cam mechanism. The movement of the cam follower (red part) is bounded inside the cavity of the cam (turquoise surfaces). The movement of the first joint is then limited by the angle of the second joint. (b) RoM of the first joint as a function of the joint corresponding to flexion/extension.

Second version The rigid body assumption is not 100% accurate but appears to work relatively well for the HiBSO. However, undesired internal DoFs exist where the tissues are not stiff enough. This phenomenon emerges mainly at the interface with the thigh. Even though the leg cuff is well tightened on the skin surface, the soft tissues enable a movement relatively to the bone. The translations and the rotations around the axis perpendicular to the femur remains limited but a large rotation about the axis that is along the femur is possible (see 3.3.2). The spherical joint therefore does not need to possess this mobility as it is redundant with the natural movement of the soft tissues.

As the leg tissues enable a non-negligible displacement, a deeper analysis of the joints used for compensating the misalignment of the main DoF may help to improve the mechanical design. Indeed if a rotational joint does not rotate more than a few degrees or if a translation of only few *mm* is required, an actual mechanical joint is not compulsory for the design to work. Such an additional joints would even downgrade the overall efficiency of the device as internal DoFs may appear.

A simple analysis was performed using a CAD model in which large misalignments between the device and hip main axes of rotation were introduced[4]. The three main movements of the hip were used and the positions of the various device joints were recorded. Fig. 5.12 presents the obtained results. As expected the first two joints correspond quite precisely to the flexion and the adduction of the hip. During adduction an important movement of the slider (i.e. the prismatic joint) is necessary. This joint is then important for the mechanism to work properly. The same remark applies to a lesser extent with flexion. The internal/external rotation requires a large rotation about one of the axes of the spherical joint. This rotation corresponds to the rotation that is naturally possible with the soft tissues of the leg. The other two axes are much less used in any of the hip movements. This suggests that the spherical joint is not crucial and may be replaced by a stiff fixation. Moreover, the rotation around the leg enabled by the soft tissues is better aligned with the axis of the femur than what the mechanical spherical joint can provide. The movements about the two other axes are then reduced even more. Actual implementation on the prototype confirms the validity of this recommendation.

5.1.2 Double Differential driven Orthosis (DDO)

Part of this subsection is adapted from (Olivier et al., 2013a) and (Olivier et al., 2014a).

Actuation

Locomotion assistance requires that the orthotic device performs cyclic trajectories. Moreover, when no assistance is required the device should be as transparent as possible in order not to hinder the natural gait movements. These two requirements are addressed by several mechanical concept as developed in chapter 4. Among the possible solutions, the double-

[4]Here again several misalignments values were tested. In the presented case the misalignment of the spherical joint at the thigh (which was identified as the most critical) is voluntarily kept important in order to better illustrate the compensation joints behavior in unfavorable cases.

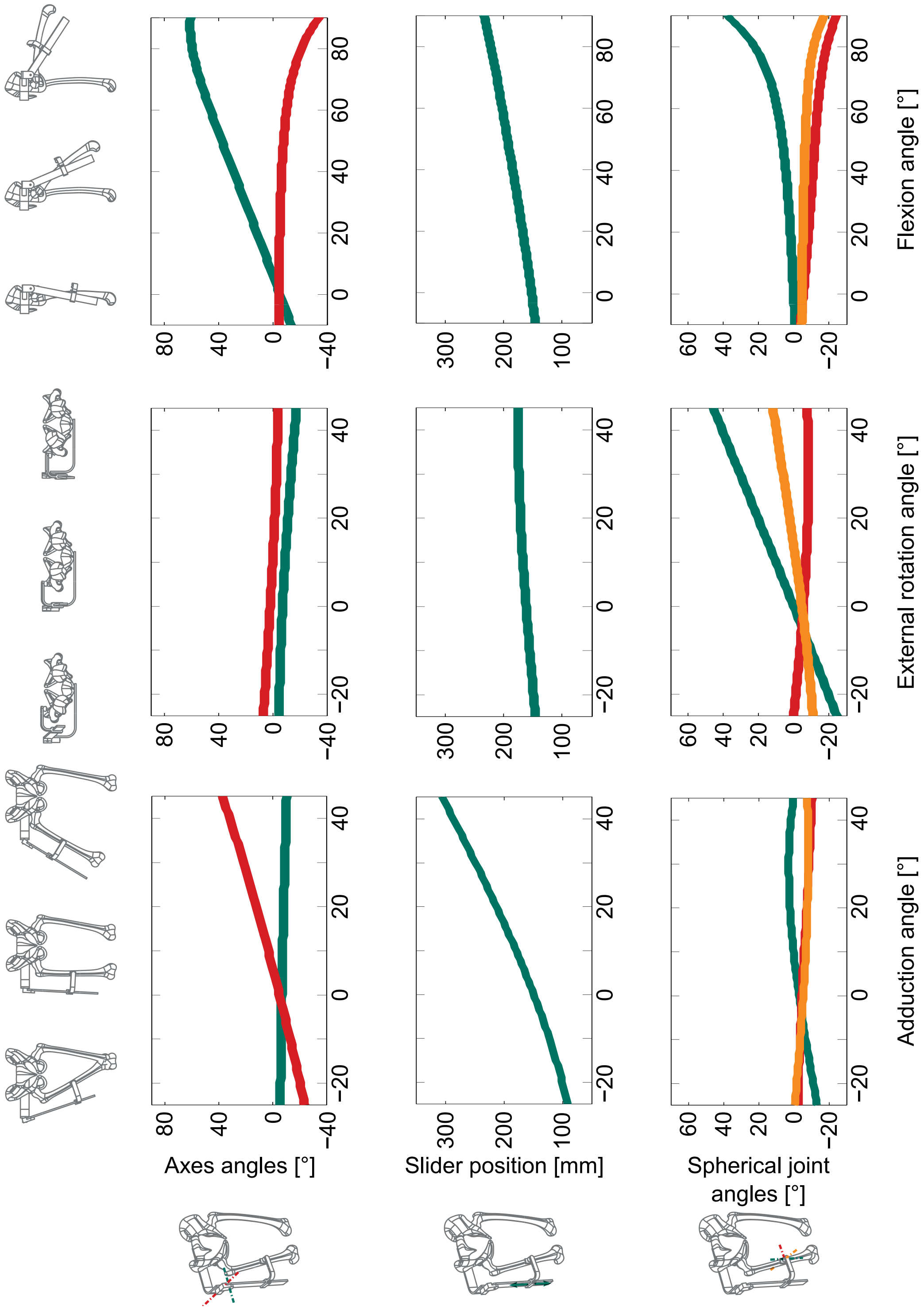


Figure 5.12 – Position of the different device joints as a function of the human movements. The prismatic joint plays an important role to compensate for the misalignments of the human and device axes of rotations. One of the DoFs of the the spherical joint is requested to move more than the other two. This rotation corresponds to the one that is naturally allowed by the rotation of the soft tissues around the femur. In the case where extreme misalignments exist, the other two rotations are also required to move significantly.

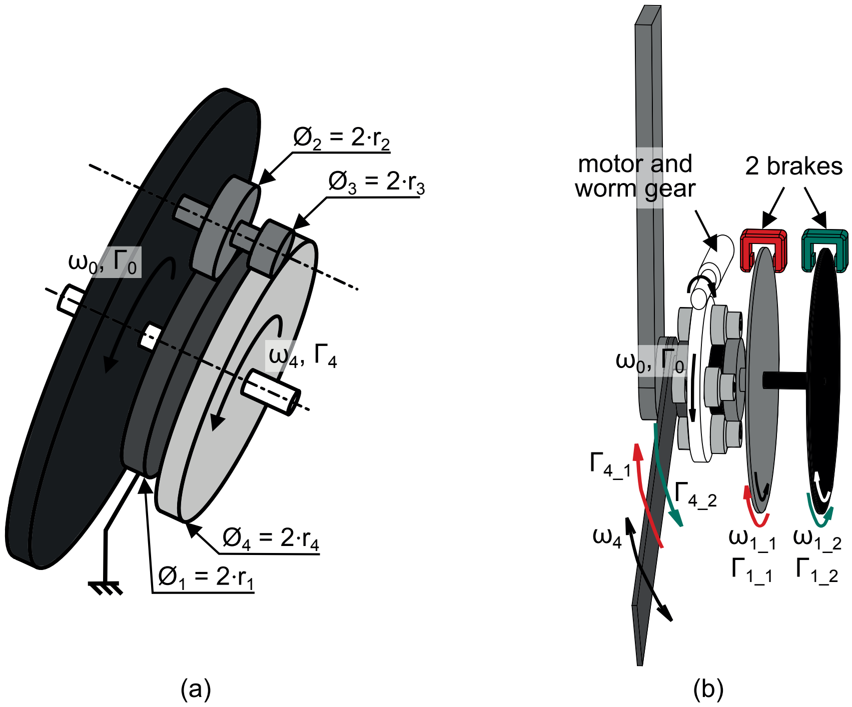


Figure 5.13 – Planetary gear train and implementation of the double-differential. (a) Planetary gear train. (b) Implementation of a double-differential mechanism based on a planetary reduction.

clutch appears promising as it represents a good tradeoff between capabilities and ease of implementation. In order to implement the clutches, we propose to use differentials controlled with brakes.

The two differentials are designed based on planetary gear trains. Such a transmission is presented on Fig. 5.13(a). One of its advantages is that it offers the possibility to have either a positive or a negative transmission ratio without changing dramatically the design. No additional gear is then necessary to reverse the movement as would be the case with a standard gear train. The transmission ratios i of a planetary gear train is given by:

$$i = \frac{\omega_0}{\omega_4} = \frac{\dot{\theta}_0}{\dot{\theta}_4} = \frac{\Gamma_4}{\Gamma_0} = \frac{r_2 \cdot r_4}{(r_1 + r_2) \cdot (r_2 - r_3)} \tag{5.29}$$

where θ_0 and θ_4 are respectively the angles of the input and the output, ω_0 and ω_4 are the corresponding angular velocities, Γ_0 and Γ_4 are the associated torques, and r_1, r_2, r_3, r_4 are the radii of the different gears as presented on Fig. 5.13(a). Note that here, it is assumed that no friction occurs. If the radius r_2 is smaller than r_3, the ratio becomes then negative. In order to create a differential mechanism from the planetary reduction gear train, the wheel number 1 needs to be able to move. In that case, the relation between the torques of one of the brakes and the output is given by:

$$\Gamma_4 = \frac{r_4}{r_3} \cdot \left(\Gamma_1 \cdot \frac{r_2}{r_1} \right) \tag{5.30}$$

One constraint remains: to be able to provide torque in both directions, the rated velocity of the motor must be greater than the ones of the outputs. The implementation of the double-differential based on planetary gear trains is shown on Fig. 5.13(b). The angular velocities of the two brakes ω_{1_1} and ω_{1_2} are influenced by the output velocity ω_4. With a sufficient input velocity ω_0, it is guaranteed that the brakes always rotate in opposite directions. Consequently torques can be provided in both directions using one brake or the other. The torque ratio between the brakes and the output should also be small enough in order to maintain the perceived inertia as low as possible. The selection of the brakes is therefore constrained by their torque capability and their weight. Bicycle disk brakes were selected for the first prototype as their torque density is very high. However, the torque they provide when controlled in open loop is not accurate enough to guarantee a precise output torque. A custom made torque sensor based on an optical sensor and deformable blades was therefore developed to enable a closed loop control. Even though this sensor demonstrated promising results, other technologies such as thick film sensors may be better suited for such measurements (Maeder et al., 2003). Future design should therefore integrate this technology.

In our design, the numerical values of the different radii presented on Fig. 5.13 are (in *mm*): $r_0 = 33$, $r_{1\_1} = 25$, $r_{2\_1} = 8$, $r_{1\_2} = 21$, $r_{2\_2} = 12$, $r_3 = 10$ and $r_4 = 23$, where the second index (*1* or *2*) of r_1 and r_2 corresponds to the associated brakes (see Fig. 5.13). The corresponding transmission ratio are then: $i_{flexion} \approx -2.1717$ and $i_{extension} \approx 4.1818$.

The velocity source ω_0 is realized by a motor and a worm gear that is used as a first stage of reduction. Its reduction ratio is *50*. Note that this first stage is non-back-drivable which is not a major problem with this mechanism. A component with a better efficiency should, however, be considered for future versions.

Kinematics

The DDO has a kinematics that is very similar to the HiBSO. The main difference is that the two first joints in the device kinematic chain are inverted. The first joint corresponds then to the flexion/extension angle. A singularity similar to the one encountered with the HiBSO is still theoretically possible. If the first joint angle reached 90°, the first joint would link to internal/external rotation. In practice, the adduction/ abduction angle is relatively limited (typically smaller than 30°). Therefore the actual workspace does not contain any singularity. Note that the adduction/abduction angle is still very sensitive to lateral movements when the flexion angle is around 90°. Here again this is due to its ambiguous definition but it does not represent any problems in use.

5.1.3 Hip orthoses comparison

The two hip orthoses presented in this section differ mainly by their actuation. The kinematics are also different but this difference is due to the length of the HiBSO actuation unit and not to a deliberate will. Moreover the difference due to these dissimilar kinematics appears only marginally in most activities.

The orthoses are therefore compared only in terms of:

- transparency
- maximal quasi-static torque that can be provided
- energy balance during a predefined dynamic trajectory

Note that this subsection is adapted from (Olivier et al., 2014a).
The transparency of an orthotic device is affected mainly by the gravity, the friction and the inertia. The centrifugal and Coriolis effects can usually be neglected. The gravity and the inertia can be modeled precisely and their effects can be largely reduced with an appropriate controller. The friction can also be modeled but as it is affected by many parameters that are difficult to master its compensation appears more challenging.
The maximal torque that can be provided depends on the transmission ratio and on the efficiency of the transmission. With the considered devices the transmission ratio depends on the position (HiBSO) or on the direction in which the torque is provided (DDO). The efficiency may also be affected by the same parameters in addition to the energy flow direction.
Eventually, in case of dynamic trajectories, energy is used by the actuation unit to make the device move. This effect needs to be considered as most of the activities that an orthotic device assists are very demanding in terms of dynamics.

Testing protocol

The three aforementioned characteristics are tested in order to compare objectively the two orthoses. The transparency and the maximal torque are assessed by performing measurements on the flexion/extension torque. In order to avoid gravitational effects, the actuated axes of the orthoses are placed vertically (see Fig. 5.14). As the friction is the most difficult effect to predict, this simplifies its analysis. The other dynamic effects are limited by performing only quasi-static trajectories (low velocities and accelerations). All the other joints of the devices are locked and no extra load is applied on them. For measuring the torque, a uni-axial force sensor is placed on the part which moves along with the leg while the other side of the joint is rigidly fixed. The torque is calculated by multiplying the distance from the joint to the force sensor by the measured force (perpendicular to the leg part). For the dynamic measurements, the motor torque is considered to be the motor torque constant multiplied by the motor input command.

Transparency As the torque due to inertia can be modeled precisely, the transparency analysis focuses on friction. Indeed the inertia of the DDO is constant and can be calculated directly from its CAD model. For the HiBSO, a detailed model is presented in 5.1.1. Different compensating conditions are tested with the two devices. The DDO is first tested without compensation and with its velocity source off. Then an active compensation is provided with

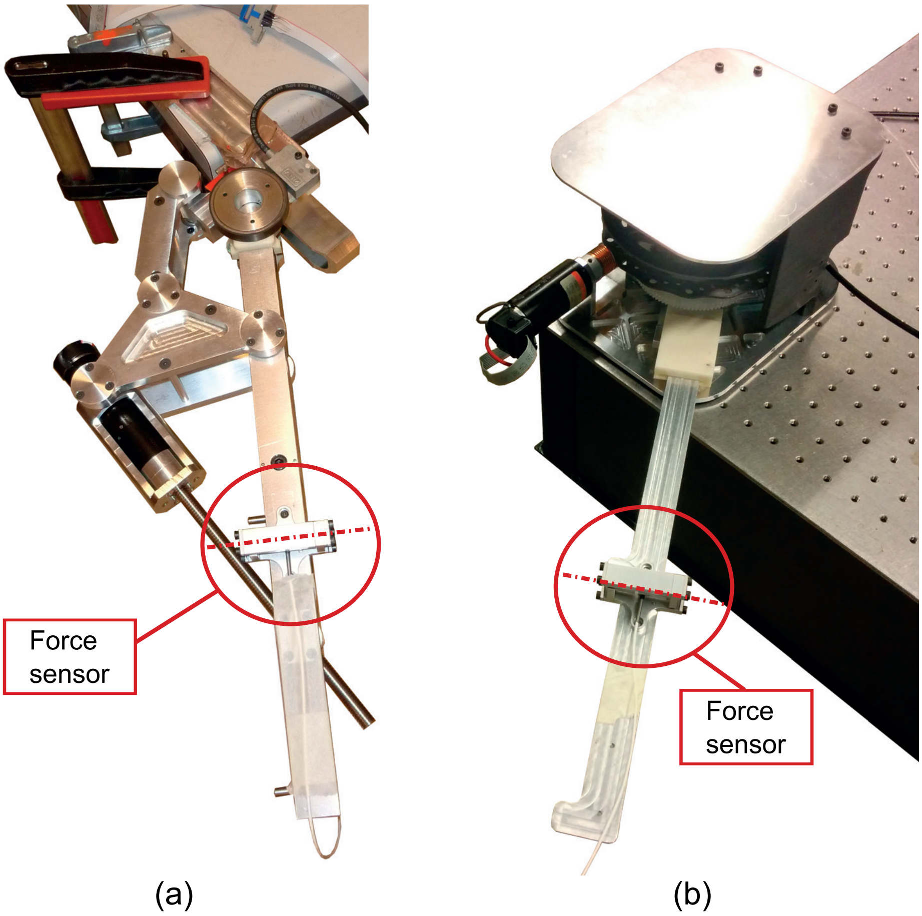


Figure 5.14 – Experimental setup for the comparison of the two hip orthoses. The orthoses are placed with their sagittal plane positioned horizontally in order to avoid any gravitational effects. The torque is measured by means of a force sensor positioned at the output. (a) HiBSO in the experimental setup. (b) DDO in the experimental setup.

the velocity source on and with one of the brakes partially active so as to compensate for the undesired friction effects (empirical tuning). The HiBSO is tested in three different conditions. The first one is without compensation (i.e. with the actuation turn off). The second one uses a simple Coulomb friction model. The friction torque is therefore expressed as:

$$\Gamma_{friction} = \Gamma_c \cdot sgn(\omega) \tag{5.31}$$

where Γ_c is a constant and ω is the rotational velocity of the mechanism. In this condition the encoder of the motor is used to estimate the output velocity ω. The applied friction compensation torque is therefore:

$$\Gamma_{friction\_compensation} = \begin{cases} -\Gamma_c & \text{if } \omega < -\omega_{min} \\ 0 & \text{if } \omega \in [-\omega_{min}; \omega_{min}] \\ \Gamma_c & \text{if } \omega > \omega_{min} \end{cases} \tag{5.32}$$

where ω_{min} is a defined threshold to cancel compensation when the velocity is close to zero. The constant Γ_c is tuned empirically so as to avoid instability while compensating for most of the friction. In the third condition, an extra-encoder placed directly on the last rotating axis is used. This enables to detect the direction of the velocity without any delay due to material deformation or play in the transmission. Moreover, dry static friction can also be compensated for as motion is detected even though the motor does not move.

Maximal quasi-static torque In order to compare the torques that the two orthoses can provide, the efficiencies of the transmissions need to be assessed. It is reminded here that the output torque is given by:

$$\Gamma_{output} = \eta_{in,out} \cdot \Gamma_{input} \cdot i \tag{5.33}$$

where Γ_{input} is the input torque, $\eta_{in,out}$ is the efficiency of the transmission and i its reduction ratio. It is then hypothesized here that the maximal torques of the two designs follow the curves presented in Fig. 5.15. Note that in equation 5.33 the output is defined relatively to the energy flow and is not necessarily the user side. Indeed when the mechanism is dissipating energy, the motor side needs to be considered as the output. i or $1/i$ evidently needs to be considered depending on this direction but $\eta_{in,out}$ may also be influenced by it.
The actuators being the same in both orthoses (Maxon RE 30 60 W), a direct and objective comparison can be done based on the ratio between the command and the measured output torques. As the transmission ratio as well as the efficiency depend on the angular position and on the direction in which the torque is applied, measurements are done on the whole workspace and in both directions. Measurements are then performed on the HiBSO when its actuator is providing 4, 6 and 10 mNm in both energy flow directions (i.e. generating and dissipating energy). With the DDO the measurements are done by activating one brake at a time (as a combined action of the two brakes would simply increase energy dissipation). The actuator being controlled as a velocity source, its torque automatically adapts to the need. The maximal transmission efficiency is then calculated with the torque ratio. As explained in 4.1.4

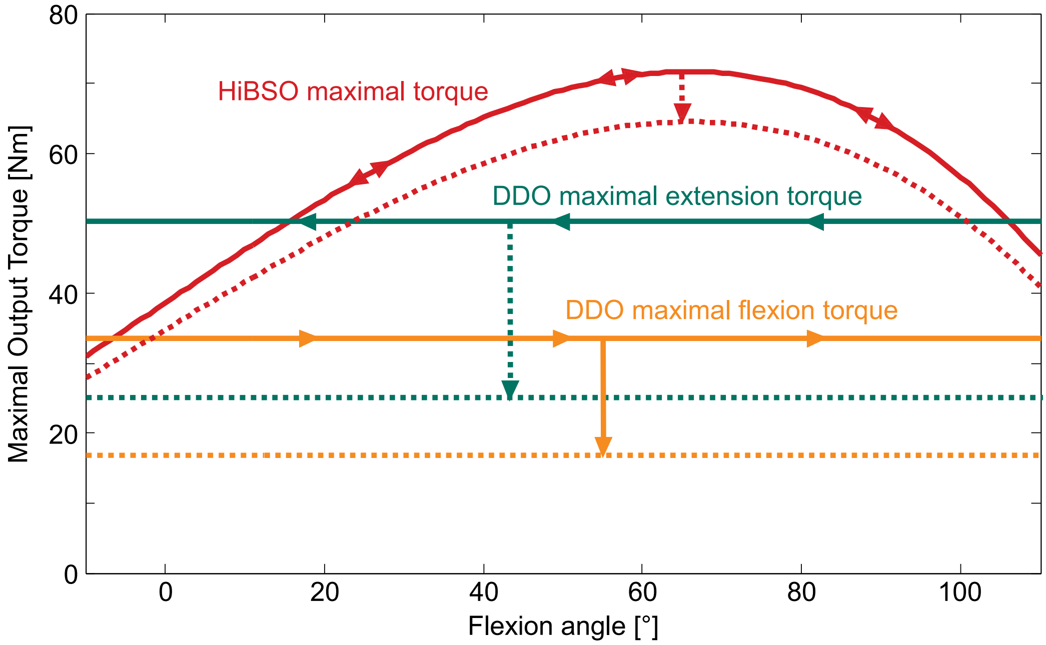


Figure 5.15 – HiBSO and DDO theoretical torques. The solid lines represent 100% efficiency and the dashed lines take into account the theoretical efficiencies of the devices. The DDO has two curves has its transmission ratio depends on the torque direction.

the velocity ratio should also be considered for the power efficiency calculation. However, only the torque ratio is of interest in this evaluation as the efficiency of the transmission (and not the clutch) is to be evaluated. Eventually, note that here only the efficiency is assessed. It is then used to estimate the maximal quasi-static torque.

Energy balance during a predefined dynamic trajectory When the mechanisms follow a trajectory, they use energy even if no additional torque is provided. The torque that the actuation needs to provide depends on all kinds of dynamic effects. The accelerations therefore play a major role. As the considered devices are essentially intended to be used for walking assistance, periodic trajectories with amplitude and frequency similar to gait trajectories are used to compare their transmission mechanisms.

As the DDO is designed to provide torque, precise position control is more delicate to perform than with the HiBSO. Therefore to be able to compare them, a trajectory is realized and recorded with the DDO and then reproduced with the HiBSO. The trajectory is realized with the double-differential mechanism by locking its two brakes alternatively. The switching between the two brakes activations is done when one of two predefined extreme positions is reached. With this particular mode of functioning the output is connected to the actuator sequentially in one direction then in the other thus moving at a constant speed in both directions (not the same as the transmission ratio is different in flexion and extension). Note that in that case the efficiency of the clutching is optimal as almost no sliding exists (the rated velocity of the output always equals the input). The changes in direction are then very

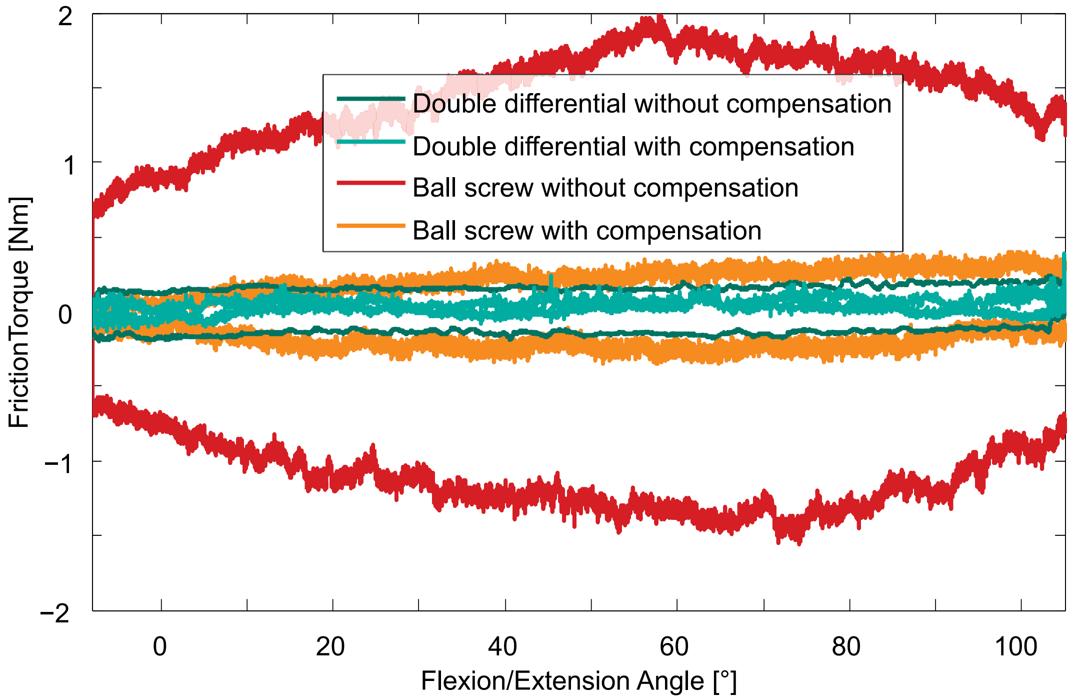


Figure 5.16 – HiBSO and DDO friction torques with and without compensation.

abrupt. For this reason smoothed trajectories are used as reference for the HiBSO. The 7th order Fourier series estimations of the trajectories appear to be an acceptable tradeoff showing decent accelerations while keeping a similar trajectory. Different cadences are evaluated so as to analyze the devices dynamic capabilities in various situations.

Results

Transparency The measurements on the resisting torque of both orthoses are presented on Fig. 5.16. The red and the dark turquoise curves show respectively the dry dynamic friction (without compensation) of the HiBSO and the doube-differential driven orthosis as a function of the position. The variable transmission ratio of the HiBSO influences greatly the resisting torque. As a consequence it varies between 0.6 $N \cdot m$ and almost 2 $N \cdot m$. The DDO presents a low resisting torque (<200 $mN \cdot m$). This low friction can be reduced even more with active compensation (see Fig. 5.16, cyan curve). The friction compensation for the HiBSO has also a very positive impact on the perceived resistive torque (see Fig. 5.16, yellow curve). However, the initiation of the movement is still problematic as the direction in which the compensation needs to be provided cannot be detected if the orthosis does not move. The effect of the dry static friction remains therefore important with the HiBSO using only the motor encoder. Fig. 5.17 shows this effect (turquoise line). With an additional encoder placed directly on the output axis the intention of the user can be better anticipated (see Fig. 5.17, yellow line).

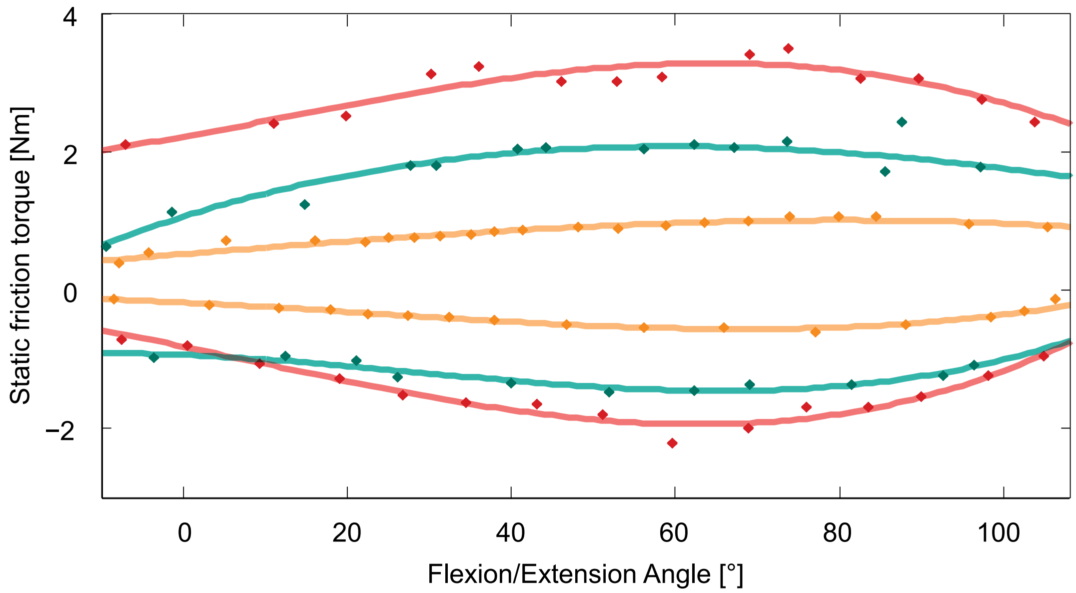


Figure 5.17 – HiBSO dry static friction. The red markers are discrete measurements on friction without compensation. The turquoise and yellow markers represent the dry static friction with compensation (using the motor and the extra encoder respectively). The red, the turquoise and the yellow lines are the fit associated with the markers of the same color.

Maximal quasi-static torque In order to estimate the efficiency of the HiBSO a simple model taking into account the transmission ratio and the motor torque is used. As it may depend on the energy flow direction two different fits are calculated. When the motor is generating mechanical work the transmission efficiency is given by:

$$\eta_{positive\ work} = \frac{\Gamma_{user}(\alpha)}{i(\alpha)\cdot\Gamma_{motor}} \tag{5.34}$$

where Γ_{user} is the torque measured on the user's side, Γ_{motor} is the torque provided by the motor and α is the mechanism output angle. When the mechanism is dissipating energy, the efficiency is calculated with:

$$\eta_{negative\ work} = \frac{i(\alpha)\cdot\Gamma_{motor}}{\Gamma_{user}(\alpha)} \tag{5.35}$$

Fig. 5.18 shows the measurements (red curves) in both cases with their associated fits (surfaces). From these fits, the efficiencies in torque generation and dissipation are estimated at 89% and 80% respectively. This confirms the high efficiency of the transmission. The significant difference between the two energy flow directions is due to the fact that the balls of the spindle drive undergo a higher pressure when the mechanism is back-driven and they

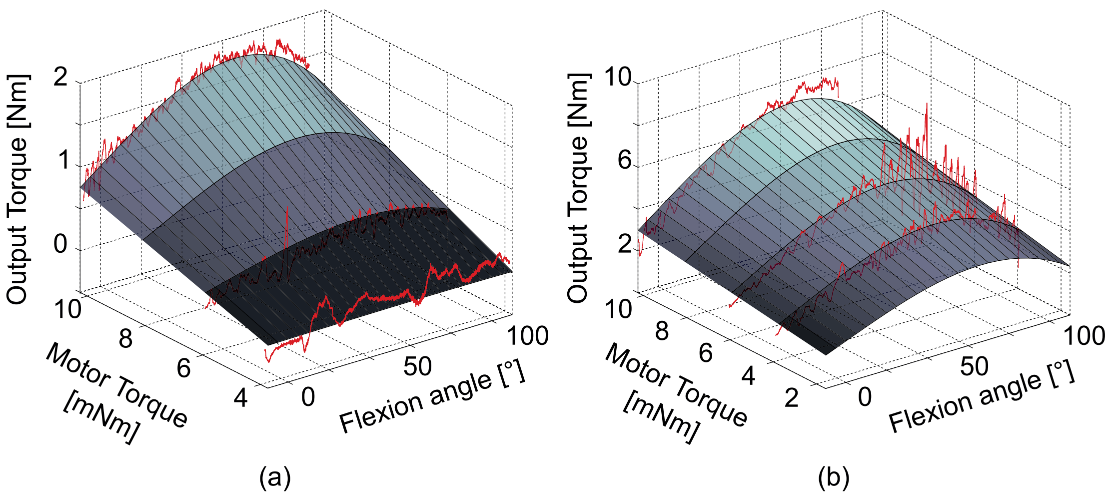


Figure 5.18 – HiBSO output torque as a function of flexion angle and motor torque. The red curves are measurements on the device and the surfaces are the fits based on equations 5.34 and 5.35. (a) Output torque when generating torque. (b) Output torque when dissipating torque.

logically generate more dissipation in that case.
The efficiency of the DDO has been measured to 43% in one direction (extension) and 39% in the other one (flexion). The overall poor efficiency is due to the worm gear in the reduction train. Fig. 5.19 shows the torque measurements in comparison with what would be obtained with an ideal transmission.

Energy balance during a predefined dynamic trajectory Different trajectories are used for comparing the dynamical capabilities of the two mechanisms. These trajectories are comparable in terms of movement amplitude but their frequencies are dissimilar so as to highlight the dynamical capabilities in various situations. As explained in the testing protocol paragraph, the trajectories are not exactly the same for the two mechanisms as the DDO is able to change direction in a very sharp manner. These trajectories are shown on Fig. 5.20 (a). The associated torques provided by the motors of both mechanisms are presented on Fig. 5.20 (b).
It can be seen that the movements when performed at low frequency do not require a lot of torque from any of the two mechanisms. However, when the frequency increases the HiBSO and the DDO react differently. Logically, the first utilizes a lot of torque to accelerate and decelerate its own inertia while the latter is not dramatically influenced. In the case of the trajectory performed at the highest cadence, the RMS torque exhibited by the HiBSO reaches 37% of its nominal capabilities. On the other hand the peak torque required by the DDO does not exceed 10% of the motor nominal torque and its RMS torque remains almost unaffected. Consequently, even though its efficiency is very low in comparison with the HiBSO's, its ability to provide torque when performing fast movements is superior.

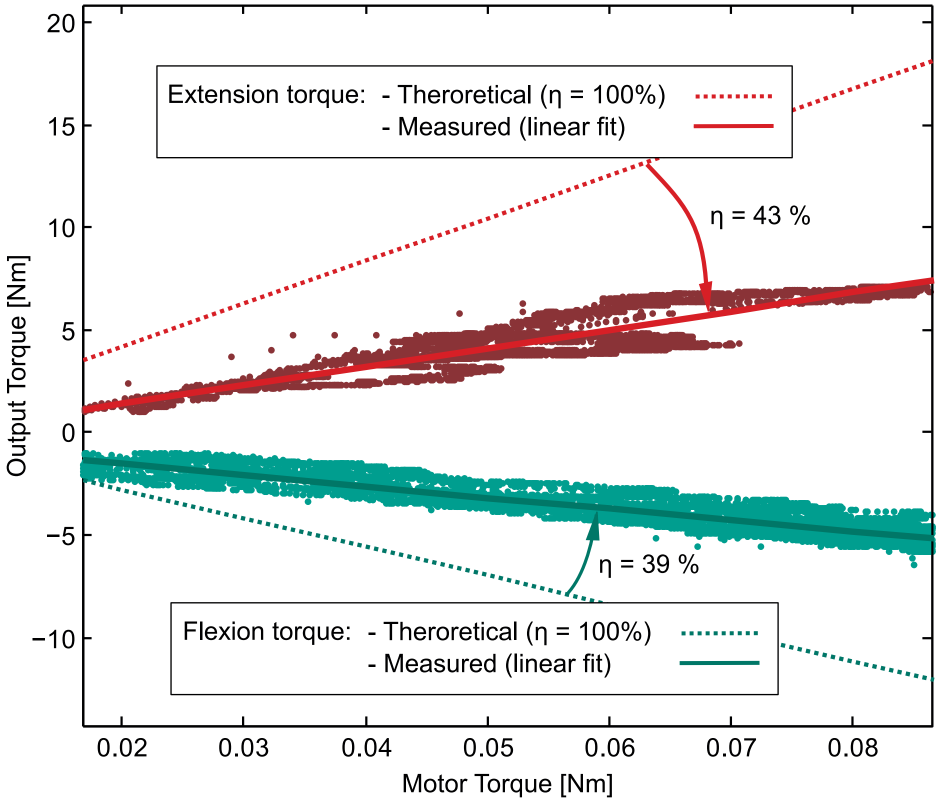


Figure 5.19 – DDO theoretical and measured flexion and extension torques.

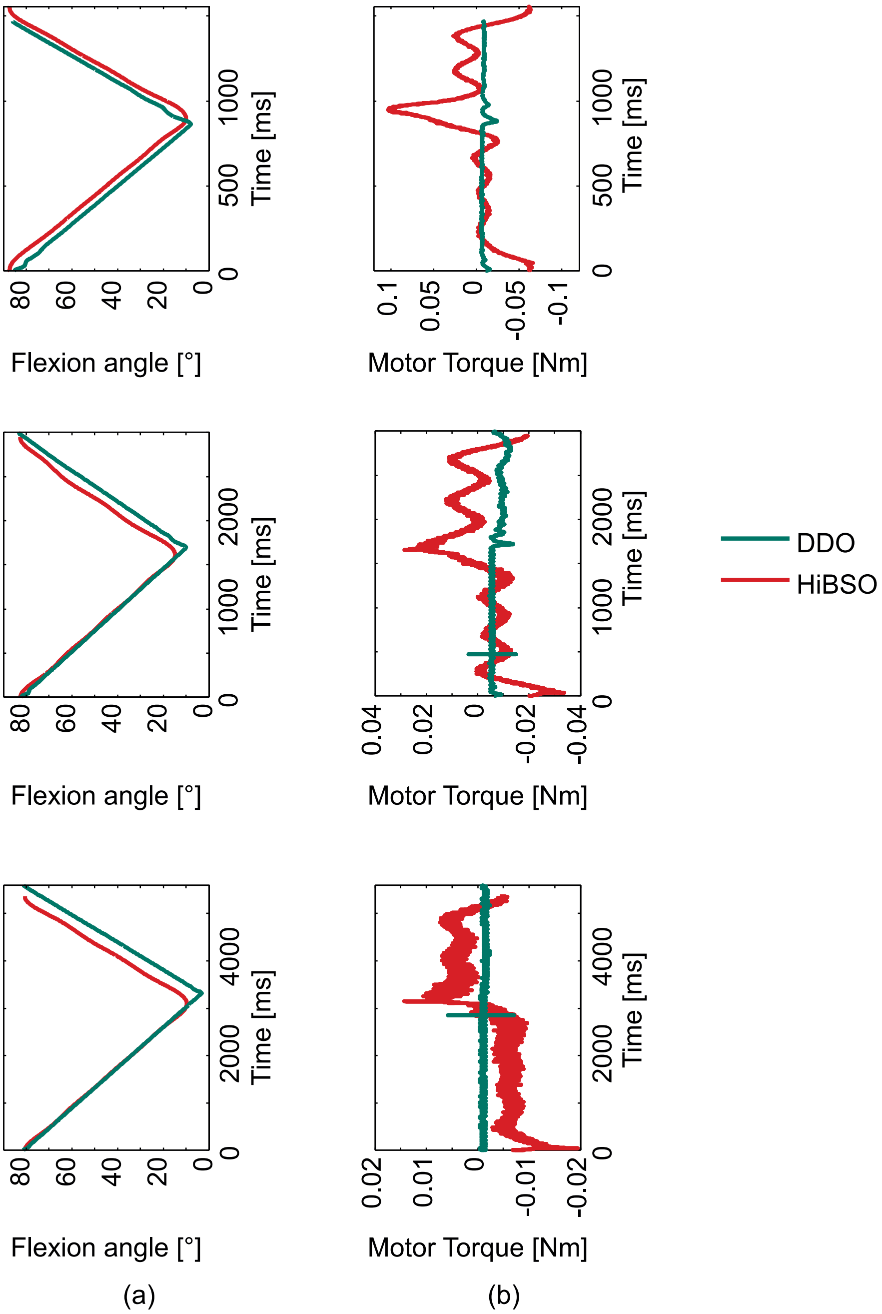


Figure 5.20 – Comparison of the dynamical capabilities of the two hip orthoses. (a) Cyclic trajectories used to compare the two orthoses. The turquoise curves are the trajectories performed with the DDO and the red ones are similar trajectories executed with the HiBSO. As the HiBSO cannot achieve very abrupt changes in direction, a 7th order Fourier series approximation of the trajectory is used as reference. (b) Associated required motor torques. Turquoise corresponds to DDO motor torque while red corresponds to the HiBSO's.

Table 5.1 – Summary of specifications and devices performances.

<table>
<tr><th rowspan="2">Orthosis</th><th colspan="2">Maximal velocity</th><th colspan="3">RMS torque for assisting walking trajectories</th><th colspan="3">Peak torque</th></tr>
<tr><th>Specification</th><th>Device performance</th><th>Specification</th><th colspan="2">Device performance</th><th>Specification</th><th colspan="2">Device performance</th></tr>
<tr><td>HiBSO</td><td rowspan="2">140 °/s</td><td>> 140 °/s (in the walking range of motion)</td><td rowspan="2">0.3 Nm/kg</td><td>~10 Nm at a pace of 90 steps/min</td><td>~6 Nm at a pace of 110 steps/min</td><td rowspan="2">~1 Nm/kg</td><td colspan="2">~70 Nm (with a flexion angle around 70°)</td></tr>
<tr><td>DDO</td><td>> 150 °/s (or unlimited if no assistance is to be provided)</td><td colspan="2">~10 Nm (does not depend on pace)</td><td>~16 Nm in flexion</td><td>25 Nm in extension</td></tr>
</table>

Discussion on hip orthoses comparison

The ideal requirements for an assistive device are difficult to address without accepting a few compromises. Indeed, a cheap and light actuator able to generate high torques and having dynamical capabilities comparable to the human joints seems difficult to achieve with the currently available technology. In this subsection, two hip orthoses with different actuation units are presented. These two distinct concepts focus on different important points of the specifications. The HiBSO uses an efficient reduction with a variable transmission ratio which enables a strong assistance for the movements that require it the most (i.e. sit-to-stand transitions) while keeping a decent dynamic in the walking RoM. On the other hand the DDO has an optimal intrinsic transparency thanks to a double clutch mechanism. Moreover its transmission ratio is different in flexion and extension in order to better mimic the typical torques that the human produce when walking. The specifications and the two designs performances are summarized in Table 5.1.

The experiments that were conducted demonstrated that the HiBSO has a very efficient transmission. The maximal static torque that it can provide is therefore very high especially in the area where its transmission ratio is high. This characteristic is helpful for stair climbing assistance of for the sit-to-strand transitions. The DDO mechanism does not offer the same versatility even with its transmission ratio that is different for flexion or extension. Indeed this characteristic is adapted to walking but is not necessarily helpful for other activities.

With the HiBSO a lot of torque is required when performing cyclic trajectories as the actuation rated inertia remains relatively large even though the transmission ratio is reduced in the walking RoM. Note that this would be the same (or even worse) with a conventional gear train with a constant reduction. The DDO presents the advantage that its actuation is always rotating at the same velocity, which eliminates the undesired inertial effects. Its poor efficiency is then largely compensated when following dynamical movements.

The transparency of the HiBSO is also inferior to what can be obtained with the DDO especially if no compensation is provided. This high resisting torque is an undesirable feature as it corresponds to what is felt by the user in case of power failure. It is reminded here that the target population is frail people who are yet able to control their legs. A highly dissipative torque (or a complete locking) may therefore have dramatic consequences (e.g. falling). With a clutch based mechanism the only consequence of a power failure is the assistance stoppage. The advantages provided by relatively simple mechanisms when compared with conventional actuation units appear really compelling. The two presented designs have advantages in various situations and it may then be envisioned that a well designed mechanism could solve most of the problems encountered with standard reduction gear trains. However, combining different mechanical concepts may potentially increasing their complexity or their weight above the acceptable limits and therefore tradeoffs are still inevitable.

5.2 Assistive knee orthosis

During normal walking the knee joint has different requirements than the hip or the ankle especially in terms of energy consumption. Indeed, the mechanical work on a complete walking cycle at the hip is slightly positive (the energy generated by the articulation is a bit more important that the energy that it dissipates). At the ankle, even more positive work is exhibited. On the contrary, the knee dissipates more energy that it generates (see 1.3). As no mechanical system has an efficiency of 100%, the excess energy that needs to be dissipated at the human articulation may be used to "assist" the assistive device in order to reduce its energy consumption.
A system incorporating a spring appears then tempting as energy can be stored without conversion so as to be returned when required. The IVT mechanism proposed in 4.1.6 represents then the ideal solution if a spring that can store enough energy is used. Theoretically, such a system used to assist walking could even operate without motor as the overall mechanical work is negative. A brake would then be used instead to dissipate the excess energy stored in the spring.
Even though the concept is appealing, the actual implementation of an IVT whose size and weight correspond to an orthotic application appears complex. An alternative is then proposed here with a double-differential mechanism. This solution uses an actuator with a reduction in series with a spring (Bertusi et al., 2014). The other end of the spring can be locked by means of a brake (ON/OFF brake). This enables to preserve the energy that is stored. Eventually, the output (part attached to the leg) is connected to this assembly through a double-differential similar the one presented in 5.1.2. The way it is controlled is however different as the input is a torque and not a velocity source. In addition to be able to recover energy from the end-effector, the spring working in combination with the double-differential prevents the motor from being overloaded in case large torques are required. Fig. 5.21 shows the proposed concept.

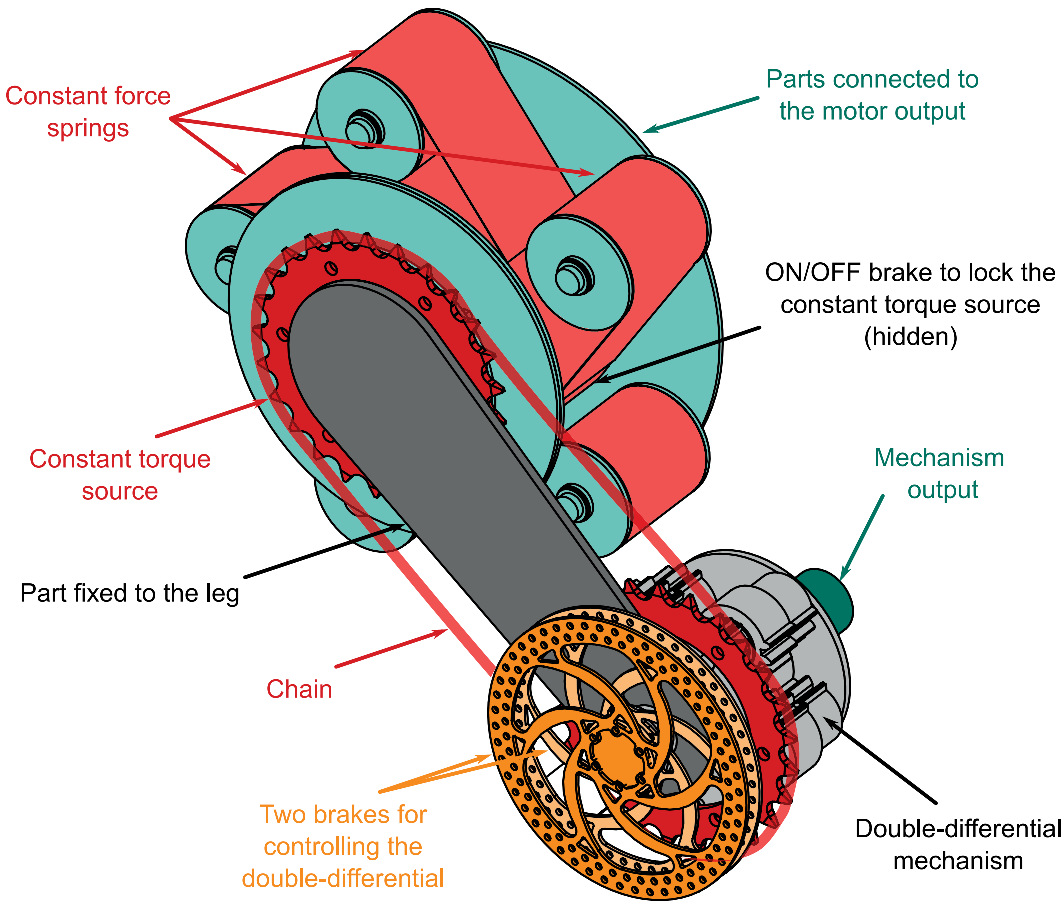


Figure 5.21 – Knee orthosis concept. Five constant force springs (red rollers on the figure) are used to provide a constant source of torque. They can be loaded either by a motor on one side or by the end-effector on the other side (if a large torque is to be absorbed). When it is required to dissipate energy with a small torque, the springs can be locked by a ON/OFF brake. This enables to preserve their stored energy when it is not used. This brake is not explicitly shown on the figure but it is used to prevent the pinion on the torque source side from moving. When the ON/OFF brake is open, the springs torque is conveyed to a double-differential mechanism through a chain. The double-differential allows to control the direction of the torque as well as its intensity.

5.2.1 Mechanical design

Elastic element

When designing a device that must store mechanical energy, the first difficulty is the elastic element. On a full walking cycle, there are typically several tens of Joules (*J*) that must be generated and a bit more that must be absorbed (see 1.3). Consequently, the spring must first be large enough to be able to store at least this amount of energy. In practice, as energy is dissipated in the brakes, even more energy is required. Second, as large reduction ratios are prohibited (so as to guarantee a good transparency), the critical factor for the selection of the elastic element is then the maximal torque it is able to produce. The system presented here is based on constant force springs[5]. Five springs are mounted on a drum. The constant force springs have one end connected to a motor (with reduction), which enables to load them. Their other ends are used as a constant torque source.

Transmission

The design concept that is presented here features a double-differential that has a transmission ratio of one third in both directions (forward and reverse). Under the assumption that there is no loss in the differentials, the torque applied by the elastic element is therefore multiplied by 3 at the output in the case where one brake is locked and the other completely open. With the same naming convention as in 5.1.2, the gears have the following radii (in *mm*): $r_0 = 30$, $r_{1\_1} = 15$, $r_{2\_1} = 15$, $r_{1\_2} = 20$, $r_{2\_2} = 10$, $r_3 = 12$ and $r_4 = 18$.
The constant torque source is connected to the double-differential mechanism by means of a chain (see Fig. 5.21).

5.2.2 Control

With this system, the torque is applied at the output by locking one of the two differentials and regulating the output torque with the brake of the second differential. It is assumed here that the springs are able to follow any trajectory imposed at the end-effector, which is a reasonable assumption as the reduction ratio is limited. The output velocity is then synchronous with the rated velocity of the springs output. In the case of the system presented in 5.1.2 the motor velocity has to be set higher than the rated output velocity and the differentials always work partially open. With springs, if the required torque is smaller than the torque that the elastic element can provide through the differential, the surplus is dissipated in the second brake (and is then lost). When generating positive mechanical work, the maximal torque is limited by the elastic element. During negative work phases, if the torque is larger than the torque provided by the springs they can be reloaded by the end-effector. The surplus torque is dissipated using

[5]Constant force springs consist of a pre-constrained blade of spring steel rolled on a drum. When unrolling this blade by pulling on its free end, a force is created. The deflection force increases up to a maximal value (typically reached when the spring is deflected 1.25 times its diameter). After that initial deflection, the force remains almost constant (reasonable changes may appear as the unrolling may modify slightly the geometry).

the second differential brake. On the other hand, if a limited torque (smaller than the torque of the elastic element) is to be dissipated, the springs should be locked in order to preserve their potential energy. This is done by using the ON/OFF brake. The negative work is obtained using any of the two differentials. Table 5.2 summarizes the different functioning modes.
The phases during which energy can be restored to the springs depend on the maximal elastic torque. The latter also defines the maximal available torque when generating positive mechanical work. With the proposed design, the maximal torque when generating energy represents about 80% of the maximal knee torque that a 70kg human develops when walking. The best tradeoff thus depends on the level of assistance that is required and on the movements that need to be assisted. With this design, energy can be sent back to the mechanism only when the torque is greater than 34.5 $N \cdot m$. With the typical trajectories and torques from (Winter, 1991) an estimation on the energy harvesting can be done. This is presented on Fig. 5.22. With the proposed design, about 1.3 J per stride cycle can be sent back to the mechanism (see the yellow area and arrows on Fig. 5.22). This energy $E_{absorbed}$ is simply calculated with:

$$E_{absorbed} = \Gamma_{spring} \cdot \Delta\alpha \tag{5.36}$$

where Γ_{spring} is the maximum torque that can be provided with the elastic elements and $\Delta\alpha$ is the variation of the end-effector angle during which the torque is greater than Γ_{spring} in the dissipation phase(s). With a softer elastic element more energy could be absorbed but in return less torque could be provided during the positive work phases as the saturation would appear earlier. For example with an elastic element able to provide 10 $N \cdot m$, about 7.5 J could be restored at each cycle (see the grey areas and dashed arrows on Fig. 5.22). Fine tuning of the mechanism may then be necessary. For example, adjusting the transmission ratios of the differentials could be beneficial in the case where assistance is required mainly in one direction (extension or flexion).

5.2.3 Discussion

Considering the drawbacks and benefits presented by the proposed mechanism leads to one reasonable conclusion: The complexity of the mechanism cannot be justified by the small amount of energy that can be saved. Indeed, if a smaller motor can technically be used with such a mechanism, the added mass due to the mechanism itself largely overpasses this small gain. The energy possibly saved is also too low to suppose that a smaller battery could be used. Moreover, the coordinated control of four active elements (one motor, one ON/OFF brake and two adjustable brakes) for only one output DoF would require a lot of tuning in order to guarantee a smooth functioning. No prototype was therefore manufactured.
Even though this development cannot be used as such, it is a first step toward new designs adapted to the specific needs of the knee joint (or more generally to orthoses able to store mechanical energy during dissipative phase of gait). Indeed, the considerations about the elastic element apply to any device intended to store energy. A clear limitation of the proposed transmission is its fixed transmission ratio. A gear box would certainly improve the energy transfer to the elastic element. A well designed IVT could perform even better. Another advan-

Table 5.2 – Knee orthosis actuation working modes. The torque can be provided in both directions regardless of the end-effector velocity. The torque that can be generated is limited by the elastic element. In modes 2 and 6, one of the differential is simply used to couple the spring with the output. When less torque is required (modes 1 and 5), the second differential is used to dissipate the excess torque. When dissipating energy, if a large torque is required (modes 4 and 8) the spring can be reloaded. One of the brakes is then locked while the other one dissipates the excess energy. If the torque is smaller than the torque of the spring (modes 3 and 7), the energy is simply dissipated using one of the two differential and the spring is locked in order to save its potential energy.

mode #	Torque direction	Velocity direction		Action on the end-effector	System control: ON/OFF brake	System control: Brake differential 1	System control: Brake differential 2
1	1	1	generate	$\Gamma_{1\_small,+}$ smaller than elastic torque	open	locked	$\Gamma_{brake2} = \Gamma_{elast} - \Gamma_{1\_small,+}$
2				$\Gamma_{1\_large,+}$ larger than elastic torque *	open	locked	open
3		2	dissipate	$\Gamma_{1\_small,-}$ smaller than elastic torque	locked	open	$\Gamma_{brake2} = \Gamma_{1\_small,-}$
4				$\Gamma_{1\_large,-}$ larger than elastic torque	open	locked	$\Gamma_{brake2} = \Gamma_{elast} - \Gamma_{1\_large,-}$
5	2	2	generate	$\Gamma_{2\_small,+}$ smaller than elastic torque	open	$\Gamma_{brake1} = \Gamma_{elast} - \Gamma_{2\_small,+}$	locked
6				$\Gamma_{2\_large,+}$ larger than elastic torque *	open	open	locked
7		1	dissipate	$\Gamma_{2\_small,-}$ smaller than elastic torque	locked	$\Gamma_{brake1} = \Gamma_{2\_small,-}$	open
8				$\Gamma_{2\_large,-}$ larger than elastic torque	open	$\Gamma_{brake1} = \Gamma_{elast} - \Gamma_{2\_large,-}$	locked

* In these cases the torque saturates at $\Gamma_{elastic}$

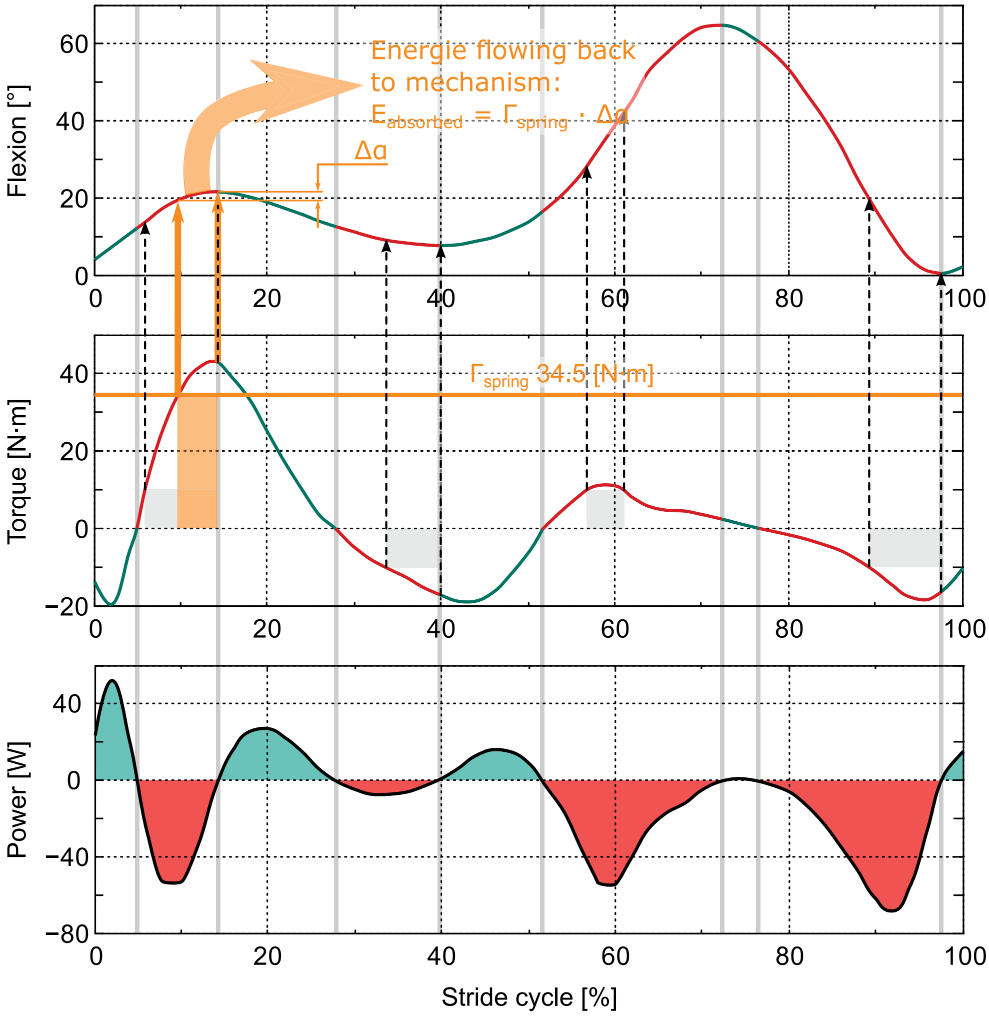


Figure 5.22 – Knee joint trajectory, torque and power during walking. The grey lines indicate the transitions between the generating and the absorbing phases. On the torque graph, the yellow line shows the elastic torque. This torque corresponds to the maximum that the device can provide during the positive work phases. It is also the minimum necessary torque able to reload the springs during the energy absorbing phases. The yellow area and arrows indicate where energy can be sent back to the mechanism with the selected springs. With softer elastic elements more energy could be sent back as illustrated by the grey areas and dashed arrows.

tage of the IVT is that no clutch mechanism would be required as a null transmission ratio[6] could be used to decouple the elastic element and the output. This would also correspond to a transparent mode.

[6]A null transmission ratio means that the torque on the elastic element side is perceived as multiplied by zero at the output. Moreover, the output velocity is transferred to the motor input (in that case, the elastic elements) with the same ratio.

6 Studies

6.1 Introduction

In the proposed studies the influence of different devices is assessed. In the first study the passive constraints imposed by the orthoses are evaluated in order to establish their potential negative influence on locomotion. No actuation is therefore used in this study.
In the second study, a motorized device is evaluated. In order to establish what is the influence of the actuation, the device is first tested with a transparent controller (i.e. the dynamic effects are compensated for by the actuator). Note that in that case the actuator adds an important load (mass, rated inertia and friction), its effect is therefore both positive and negative. In a second part, an assistive controller is evaluated so as to establish if the negative effects of a wearable device can be exceeded by the benefits proposed by an actuated system.

6.2 Passive influence of a wearable device on gait

Wearing an orthotic device has unquestionably an effect on the user's movements. Before considering any support strategies, the passive impact of the device needs to be well understood. Indeed it is important to make sure that the negative impact of an exoskeleton does not outbalance its benefits.
Even though the orthosis can be kinematically well designed, tailored on its final user and made of light materials, additional loads and constraints appear inexorably. These new forces may then make the human movements more physically demanding, reduce the original articulations Range of Motion (RoM) or simply modify the usual motion trajectories.
This study aims therefore at showing how healthy subjects are influenced by an orthotic device when they are walking.

6.2.1 Potential sources of movement alteration

Increased load

An exoskeleton is a by definition a rigid articulated structure located outside of the body. The first source of movement alteration comes from the modification of the system dynamics. The exoskeleton rigid structure has moving masses which create inertial and gravitational forces. In addition, it is usually made of several parts that move relatively to each other. Friction and damping may then appear between the moving parts. In some cases elastic elements may have a non-negligible influence on the movement. Consequently this system's dynamics can by expressed as any multi-body system with the well known equation:

$$\Gamma = I(q) \cdot \ddot{q} + G(q) + C(q, \dot{q}) \cdot \dot{q} + F(q, \dot{q}) + K(q) \tag{6.1}$$

where Γ is the generalized torques and force vector, q is the generalized displacement vector, I is the inertia matrix, G is the the vector expressing the gravitational effects, C corresponds to the centrifugal and Coriolis effect, F corresponds to the friction and damping effects and K is the vector containing the stiffness terms.

Kinematic constraints

The second source of movement alteration comes from the modification of kinematics. If the design of an exoskeleton does not encompass some mobilities, the human who is wearing it may see his/her own original mobility restrained (if the exoskeleton is stiff enough and firmly attached to the body). It is the case with the Ekso and the ReWalk notably as these devices limit the legs workspace to the sagittal plane only.
Moreover, as an exoskeleton creates one or more kinematic loops when attached to a wearer, it is possible that the human mobility is reduced even if the exoskeleton is designed to enable all kinds of movements[1]. As explained in 4.2 two types of kinematics can be considered: the over-constrained systems and systems with additional DoFs.
The mobility reduction may also be due to the mechanical stops of some of the structure joints. In that case the locking of a DoF is local. It is therefore a reduction of the RoM and is usually not an issue in the walking RoM. In practice no DoF is ever fully locked as the interface is always made through soft tissues and the structures are never perfectly stiff. In that case the constraints can also be expressed with equation 6.1.

Discomfort and constraints due to the interfaces

The last hypothetical source of movement alteration is the interface with the wearer. Indeed, the mechanical structure of the exoskeleton cannot be attached directly on the user. Cuffs, straps, molded parts or a combination of these are therefore usually used to maintain the

[1] It is important to consider the human mobility and not the overall mobility as the latter may contain Degrees of Freedom (DoFs) that concern only the exoskeleton (i.e. internal DoFs).

mechanical and the biological parts attached together.
These interfaces are crucial as they must transmit the potentially high efforts between the two sub-systems (body and exoskeleton) while remaining comfortable. As they are in direct contact with the user they may generate discomfort when moving as they compress soft tissues or when they rub on the skin. The possible ache may then lead to adaptations which compromise the original execution of trajectories.
Moreover in the case of large interfaces (required for instance when high efforts need to be transferred between the device and the user) other joints may be affected. This is particularly noticeable at the trunk level. Indeed the trunk is not rigid as the pelvis and the vertebrae move relatively to each other during walking (Crosbie et al., 1997). A large number of DoFs may then be affected when wearing an interface made of relatively large and rigid parts. On the other hand, the interface at a non-articulated segment (such as the thigh) is not as critical from this point of view as the segment is composed of one bone only (i.e. the femur in the case of the thigh) and no internal mobility can be affected.

6.2.2 Existing studies on the influence of passive devices

Several research groups work on the interaction between exoskeletons on human beings. Studies therefore exist on the passive influence of a robotic structure on a human. Note that here the compensation strategies that may be implemented in the control of these devices is not considered. The present subsection is divided according to the three sources of movement alteration described in 6.2.1.

Effect of the segments dynamics

Gait being a dynamic activity, it is logically affected by any mechanical impedance changes (inertia, damping, stiffness). Some groups have studied the influence of such parameters so as to determine the maximal acceptable load that can be applied on subjects while they are walking.
The influence of mass added on the subject's pelvis or ankles was studied in (Meuleman et al., 2013). In that study the extra load was compensated for with a custom body-weight support. The authors suggest that this replicates the effect of inertia only. Different conditions (with different masses placed at various location of the subject's body) were thus assessed (by recording gait kinematic, Electromyography (EMG) and metabolic rate data). Their goal was to determine the maximal acceptable inertia for gait training exoskeletons. Meuleman demonstrated that adding inertia at the ankle has an effect on the hip flexion RoM as well as at the upper body level. However, this effect was considered negligible for masses lighter than 2 kg as its influence remains within the inter subjects variability. In (Browning et al., 2007), a similar study is proposed. The difference with Meuleman's study is that Browning et al. did not compensate for the extra load with any body-weight support. Their results were however very similar and here again no kinematic changes could be clearly identified for masses lighter

than 2 *kg*.

Effect of the kinematics configuration

The effect of the kinematics configuration has been studied more intensively for the upper limb than for the lower extremities. However, a recent study from Zanotto addresses the problem of the misalignment of a mechanism acting on the knee joint (Zanotto et al., 2015). In this study the focus is given on the influence of a mismatch between the lengths of the segments of a gait rehabilitation robot and that of its user. The interaction forces were measured during walking for different values of knee misalignment. It was highlighted that poorly adjusted segments tend to augment significantly the interaction forces.
Another evidence that the effect of the kinematic design is important was discovered by Rossi et al. when studying the influence of added mass at the knee level (Rossi et al., 2013). To do so, they placed different masses between 0.5 and 2.5 *kg* on a knee orthosis and recorded various gait parameters. They demonstrated that no influence could be clearly attributed to the extra mass. However, the orthosis had a pronounced effect on gait when compared with the baseline (without knee orthosis). The undesired mechanical constraint due to the orthosis therefore exhibited a more severe gait alteration than the mass, which strongly suggests that this point cannot be overlooked when designing a gait training robot or an exoskeleton.
Most of the commercially available exoskeletons have only two active joints per leg (at the hip and knee joints). Even though the ankle is not critical for achieving the locomotion task (Veneman et al., 2007) the movements and power generated at this level are important in normal walking (see 1.3). The effect of ankle locking was therefore studied in (Olivier et al., 2015). Note that an adaptation of this study is also presented in appendix B.

Effect of the interface with the user

To our knowledge the isolated effect of an exoskeleton interface on gait has never been studied. Only a few studies on the effect of comparable rigid structures placed on the trunk have been found. Willems for instance investigated the effect of a plaster cast on the lumbosacral joint motion (Willems et al., 1997). So called Thoracolumbosacral orthosis effects have also been assessed for instance in (vander Kooi et al., 2004). However, even though these devices exhibit similarities with some exoskeleton trunk interfaces, differences exist and the purpose of the studies are not exactly the same. The obtained results are therefore difficult to adapt to another context. The gait perturbations due to such interfaces are consequently not well mastered.

6.2.3 Methods

Task

In order to identify the effects of the various factors that may influence gait, several conditions are compared. The kinematic variations induced by different conditions are therefore recorded

on several subjects walking on a treadmill. The first recording is always a baseline measurement in order to enable the subjects to get used to the experimental setup. The velocity of the treadmill is self-selected prior to any recording and is kept identical for all conditions. The different conditions are recorded on the different subjects with a randomized order in order to avoid possible effects induced by external factors such as fatigue or adaptation to the treadmill walking.
The influence of the interface or of the mass only can relatively easily be isolated even though some side effects are inevitable. Indeed an additional mass needs to be attached with some kind of interface and any interface has a mass. On the other hand, the constraints induced by the kinematics cannot be considered independently for two reasons. First, no robotic orthotic device possesses a negligible mass distribution and second they unquestionably need to be attached to the body through a relatively robust interface. Consequently this factor is considered in different configurations (i.e. different kinematic organizations with different numbers of DoFs) but inevitably in addition to the two other factors. Note that the effect of friction and viscosity is considered negligible here as no actuation unit is present during the experiment.
Six conditions are therefore compared:

- Baseline
- Additional mass
- Interface only
- Kinematics 1: Hip Ball-Screw Orthosis (HiBSO)
- Kinematics 2: Hip module of the lower limb exoskeleton research platform with unlocked flexion/extension
- Kinematics 3: Hip module of the lower limb exoskeleton research platform with unlocked flexion/extension and adduction/abduction

Details about the conditions are provided in the "Material" part below.

Subjects

Four young and healthy male subjects participated in the study. Table 6.1 shows their main characteristics.

Material

Motion capture The gait trajectories are recorded with a motion capture system based on passive markers (OptiTrack system, NaturalPoint Inc, *USA*). The system works with twelve

Table 6.1 – Passive influence of hip orthoses experiment subjects' main characteristics.

Subject	Gender	Age (years)	Height (cm)	Weight (kg)
S1	Male	31	175	72
S2	Male	31	185	84
S3	Male	26	202	100
S4	Male	31	185	77

cameras, which considerably limits the possibilities of occlusion. The frequency of the acquisition is 200 *Hz*. Depending on the conditions, a set of 33 to 34 markers is placed on the subjects so as to be able to track the different body segments[2].

Hip orthoses Two different hip orthoses are used to compare the influence of different kinematic configurations. The first one is the HiBSO (see 5.1). In order to be able to better assess the influence of the three factors mentioned above, the actuation units were removed. This considerably limits the friction and inertia effects induced by the actuators and their transmissions. It also partially limits the device weight and makes it more similar to the other orthosis.

The second hip orthosis is the hip module of the lower limb exoskeleton research platform described in (Bartenbach et al., 2015a) (see Fig. 6.1). The most important feature of this device is that the RoM of one or several DoFs can be reduced (or even completely locked). While the HiBSO is based on a kinematics with additional DoFs[3], the second device requires a precise alignment of the different centers of rotation (see 4.2.1 and Fig. 4.6). For this study, two different configurations are used. The first one uses two DoFs corresponding to flexion/extension and adduction/abduction. The internal/external rotation is not required as the soft tissues on the thigh already enable this movement. This DoF would then be redundant and therefore create an internal mobility, which may be uncomfortable for the wearer. Note that this DoF can be unlocked in the case where the knee module is used [4]. In the second configuration, the adduction/abduction is locked and thus only the flexion/extension is possible.

The weight and the mass distribution of the two devices are quasi-identical. The HiBSO weighs

[2]Depending on which interface is used in the tested condition, one or two markers are placed on the lower back so as to limit the possible occlusions. Four additional markers are also used during the initialization so as to properly localized the ankle and knee joints.

[3]The newest version of the HiBSO has in fact only three mechanical DoFs (see 5.1). The other three are provided by soft tissues deformation at the thigh level. The kinematic analysis demonstrated that mostly the rotation around the femur is required, while the other two move only marginally (see Fig. 5.12). The compliance of the tissues around the femur is perfectly acceptable and the required angles can be easily achieved without creating high reaction forces.

[4]With the knee module the third DoF at the hip level is required as the cuff under the knee needs to be able to rotate with the leg. In the initial position (straight leg, no internal/external rotation and no adduction/abduction) the knee axis is co-localized with the axis of the orthosis knee module. With the rotation around the femur, this co-localization could not persist and the flexion of the knee would therefore be impossible.

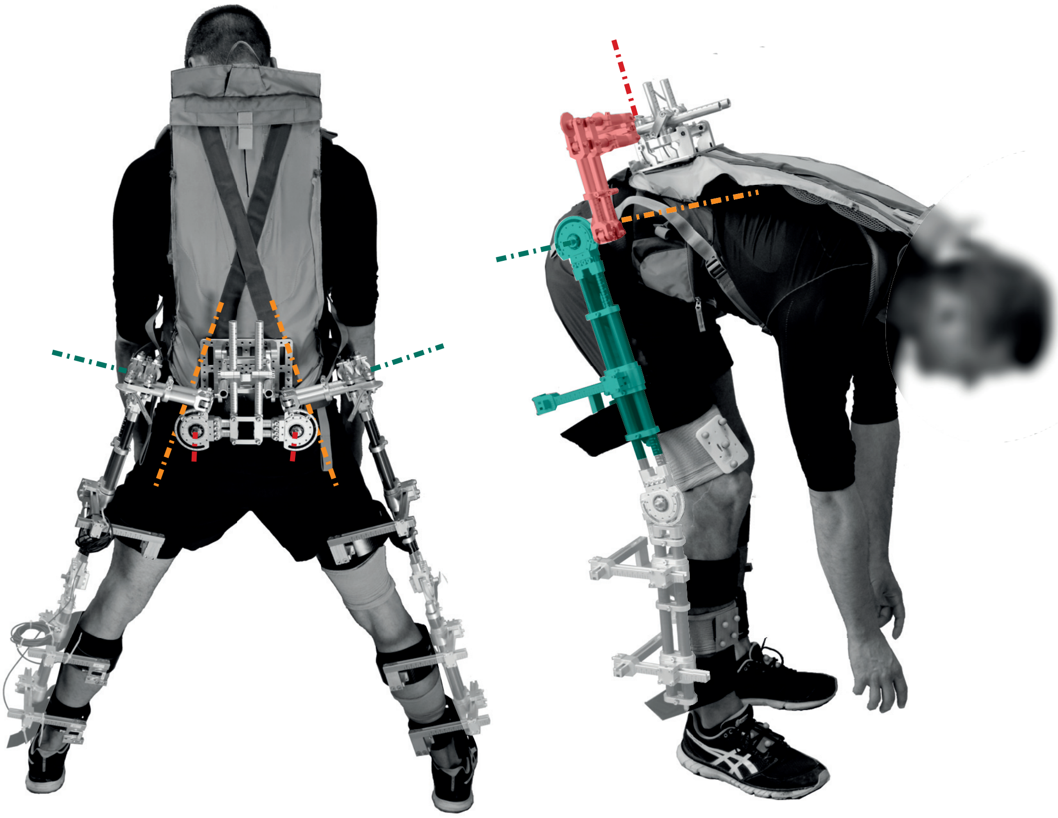


Figure 6.1 – Lower limb exoskeleton research platform (Sensory-Motor Systems Lab – ETHZ). Only the hip module is used for this study. The parts that are not used are displayed in lighter shades. The three axes of rotation of the hip module are shown in color. In this study only the flexion/extension (turquoise) and the adduction/abduction (red) axes are used; the yellow axis corresponds to internal/external rotation and is not required when only the hip module is used. On the right, the parts displayed in red move only when adduction/abduction is performed. The turquoise parts simply follow the thigh and therefore move with adduction/abduction but also with flexion/extension.

5.6 *kg* from which 1.7 *kg* move with each leg. The second hip orthosis weighs 6.7 *kg*. The parts moving with the legs represent between 1.3 *kg* for the flexion/extension and 2.4 *kg* for the adduction/abduction. This difference comes from the fact that there is less parts in motion when the legs move in flexion/extension than in adduction/abduction (see Fig. 6.1).
In addition to the kinematics, another significant difference exists between the two devices: the interfaces with the wearer. The HiBSO being intended to provide relatively large torques, its cuffs have relatively large surfaces in contact with the user (this interface is further described below, see paragraph "Orthosis interface"). The second hip orthosis on the other hand has relatively narrow cuffs at the thighs. Its upper body interface is simply a backpack with straps at the pelvis and torso levels.

Additional mass The condition with additional mass only is realized by using short pants provided with pockets in which weights can be inserted. The pockets are located at the pelvis and at the thigh levels. The mass distribution was therefore set to mimic the one of the HiBSO (i.e. 5.6 *kg* including 1.7 *kg* placed on each leg). The short pants being made of relatively thick fabric, they may restrain large amplitude movements. However, in the walking RoM it is assumed that this influence is negligible.

Orthosis interface The thigh and trunk interfaces of the HiBSO have large and rigid surfaces in contact with the body. It is thus hypothesized that they are more constraining than the compact cuffs and the backpack which compose the interface of the second orthosis. Therefore the three HiBSO molded parts (two for the thighs and one for the trunk) and their straps are used so as to test the effect of the interface only. Evidently this interface has a non-null mass which represents approximately 30% of the complete hip orthosis. The effect of this limited mass is considered to be relatively unimportant in comparison with the ones of the additional mass and the hip orthoses conditions.

6.2.4 Results

In order to detect the global effect of the different conditions, the data from all subjects were combined. Fig. 6.2 shows the hip, knee and ankle trajectories in the sagittal plane for the six different conditions. The curves appear to be very similar to each other and are all within the inter-subject variability[5]. The most important variation is an offset of a few degrees that can be observed in the additional mass condition on the hip flexion/extension sub-figure. This offset could be interpreted as a postural change, for instance to compensate for the modification of the Center of Mass (CoM) location. However, it is more probable that this offset is due to the initialization of the markers as similar differences can be observed between the conditions recorded on individual subjects.
The cadence is not influenced dramatically by the different conditions[6]. The largest difference

[5]No standard deviation or confidence interval is displayed on Fig. 6.2 so as to preserve a decent readability.
[6]As the velocity is constant, this parameter is directly linked with the stride length.

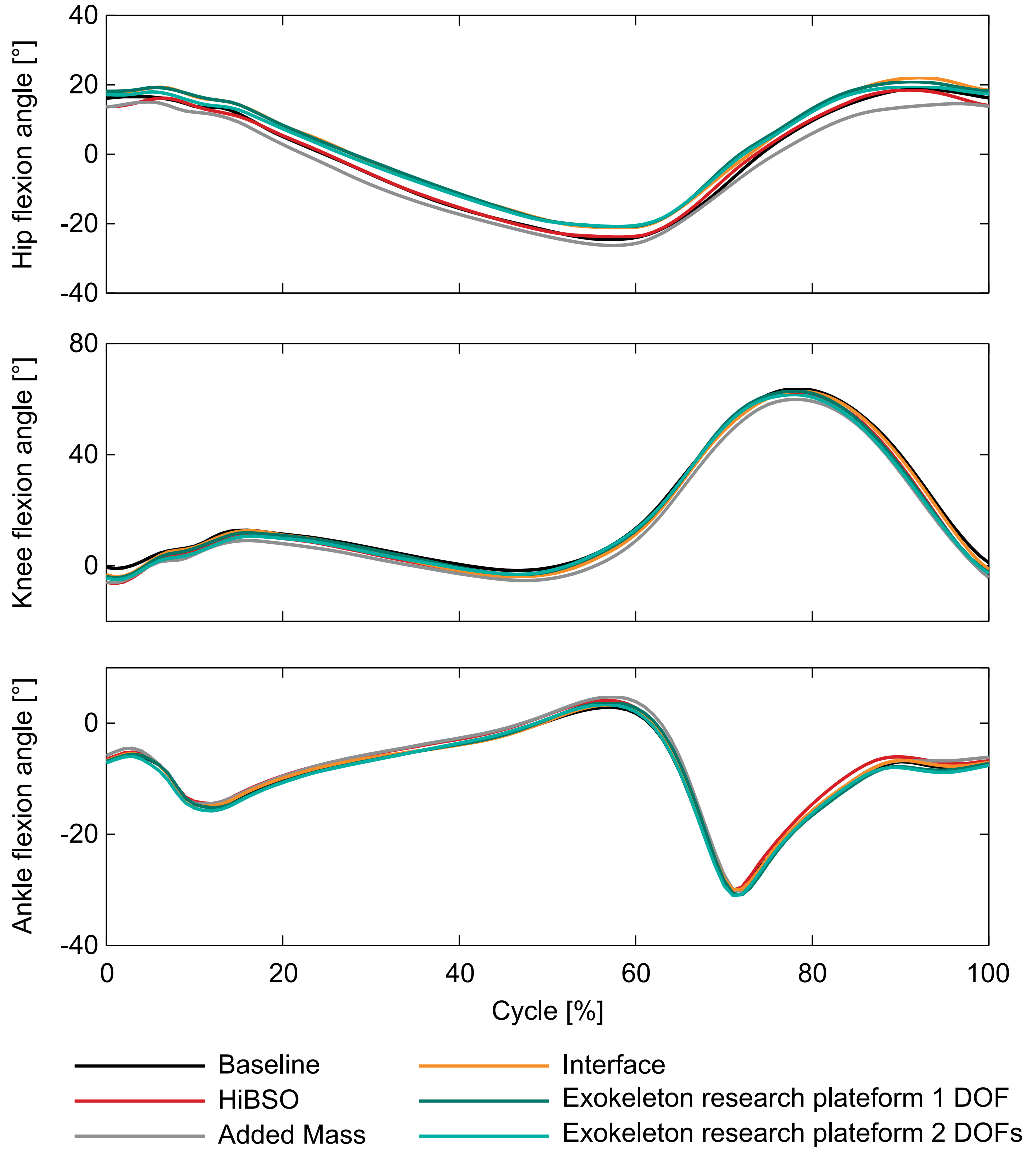


Figure 6.2 – Hip, knee and ankle trajectories for the six different conditions. The cycle starts at Heel Strike (HS). The colors corresponding to the different conditions are indicated on the figure itself. An important overlapping of the different curves can be observed at the three joints.

is induced by the lower limb exoskeleton research platform in its two configurations. This difference remains very limited as the time to perform a full stride augments of approximately 16 *ms* in average, which corresponds to a cadence difference of 1.4 *step*/*min*.
The step height is relatively constant in all the conditions. The largest observed difference is 9 *mm* for subjects S2 and S4 in conditions "interface" and "lower limb exoskeleton research platform with two DoFs" respectively.
The RoM of the different joints exhibit maximal differences of 2.7°, 2.8° and 2.4° respectively for the hip, the knee and the ankle in the sagittal plane. The individual data of the subjects reveal larger differences (typically maximum 5° for all joints). One subject (subject S2) had his knee RoM more influenced in the condition where the HiBSO was worn (increase of approximately 10°). Subject S1 and S4 also had an increase in their knee RoM but less important (around 3°). Subject S3 on the other hand had a RoM reduction of 2°.
The adduction/abduction angles are also affected by the different conditions in a limited manner. At the hip, subjects S1, S2 and S4 had variations of less that 3°. Subject S2 had a reduction of about 7° in all the non-baseline conditions excepted the additional mass one.

6.2.5 Discussion

In the context of walking on a treadmill while being imposed *a priori* hindering conditions, only negligible alterations can be observed at the joints kinematic level. The standard gait parameters (notably cadence, stride length, step height) do not appear to be affected either. The more important differences that can be observed when considering the data of individual subjects remain limited and do not appear to exhibit any trend (i.e. when some subjects tend to augment one parameter, others may diminish or keep it stable). Moreover, most of the differences induced by the different conditions are within the intra-subject standard deviation. One could then argue that the different conditions that were tested do not influence the wearer and therefore are acceptable for an assistive device. This statement should however be moderated for several reasons.
The first one comes from a clear limitation of our study: only four subjects were evaluated. With this number of subjects it is impossible to make sure that general trends are detected. Indeed, individual data demonstrated small but significant differences between the conditions. These differences are not necessarily consistent between subjects. Therefore it is possible that different adaptation strategies are adopted by groups of people. Evidently such groups cannot be identified with four subjects.
In addition, all the subjects who participated in the study are young, healthy and have regular physical activity. Their aptitude to reject perturbation is consequently optimal. Generating more torque to follow the planed trajectory is therefore not a problem. A more detailed analysis at the joints torques level may reveal that the torque patterns are modified.
The third reason comes from the fact that the human beings can adapt to unexpected conditions by stiffening their articulations (co-contraction). In that case, an error induced by an external factor is rejected as it would be with a proportional controller with a high gain. The torque remains then identical as long as no perturbation occurs but the metabolic cost of the

subject increases due to the intensified co-contraction. It is then probable that the different conditions are perceived as perturbations and therefore rejected by this augmentation in the muscular activity. Metabolic rate and muscular activity measurements would be required to confirm this hypothesis.
Even in the case where the different conditions would not influence at all treadmill walking, additional investigations are required for determining their impact on other activities. Indeed, walking on a straight line (as is the case on a treadmill) does not require much activity taking place out of the sagittal plane. Consequently, conditions like "lower limb exoskeleton research platform with unlocked flexion/extension" are not really dramatically constraining as the adduction/abduction is anyway unimportant. Nevertheless, it is supposed that walking on a curved line or on a inclined surface would be more challenging in that condition. Similar comments apply to the interface of the HiBSO. Its trunk interface is relatively rigid and therefore it restrains the movements of the spine. Simple activities like standing up from a sitting position would then probably be affected by the conditions where this interface is used.
Even though the study has limitations, it appears that the perturbations induced by the various conditions have a moderated impact on the tested activity (i.e. walking on a treadmill or more generally walking on a straight line). The RoM is not limited and the possible perturbation are almost completely rejected. For simple activities the imposed level of constraint is then acceptable at least at the movements amplitudes and trajectories levels.
This study also partially explains why the commercially available exoskeletons with limited DoFs (Ekso, ReWalk, Indeego, ...) perform well when mobilizing patients legs to enable walking on a straight line. Indeed, the level of constraint induced by the kinematics and the interfaces[7] is relatively high but does not prevent the user to execute the movements that enables forward progression.

6.3 Influence of an assistive hip orthosis on gait

In this study the influence of an assistive controller implemented on the HiBSO is tested. The subjects are therefore asked to perform a walking task on a treadmill at an imposed velocity. In order to have a baseline, the subjects walk first without the device. In a second phase, a transparent controller is implemented on the assistive orthosis in order to measure the impact of the device without assistance. Note that this condition is similar to the study presented in 6.2 with the difference that the actuator is present and used to compensate the dynamic effect of the orthosis. As the actuator is responsible for a large part of this undesired effect, this condition is *a priori* less favorable than the condition without actuator. With the third condition assistance is provided to the user. Eventually, a fourth measurement is performed in order to detect any after effects which may be due to an adaptation to assistance or simply to fatigue.
In the next two sub-sections the control modes are introduced. The recording method is then

[7] In the case of these devices, the last parameter (i.e. the additional load due to mass, inertia, friction, etc.) is not critical as it is only perceived by the device. Indeed, these devices being intended for mobilization, mechanical transparency is not useful as the user is not supposed to actively participate to the movement.

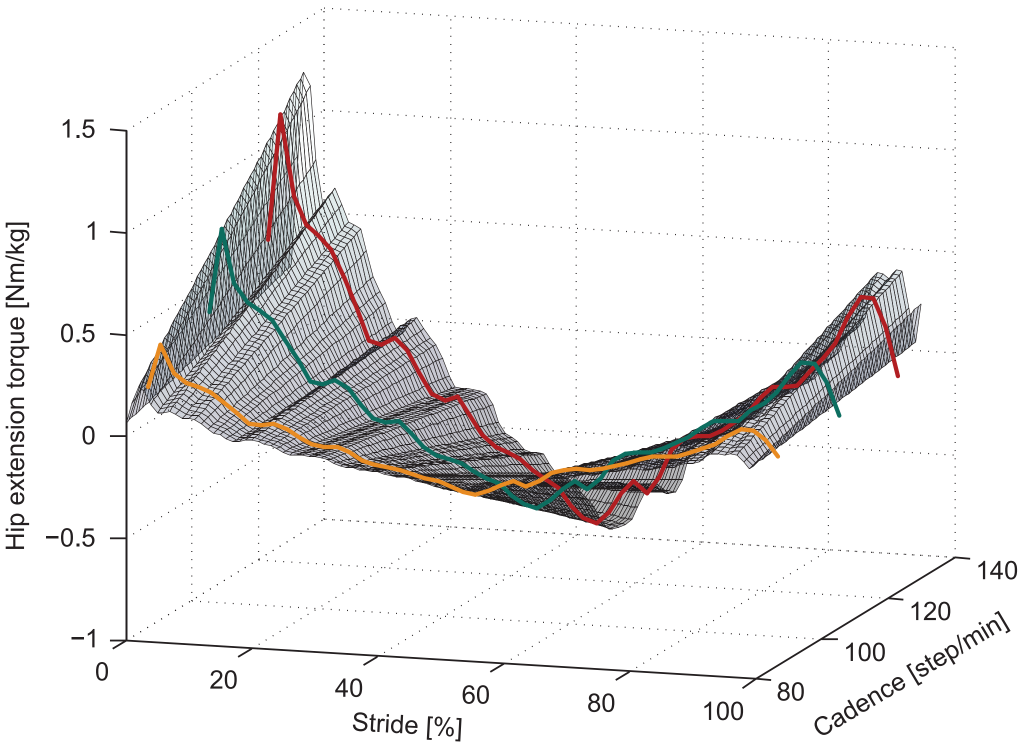


Figure 6.3 – Nominal hip extension torque at different cadences. The curves are adapted from (Winter, 1991). The yellow curve corresponds to a cadence of 86.8 *steps*/ *min*, the turquoise and the red ones to 105.3 and 123.1 *steps*/ *min* respectively. The surface is the fit according to equation 6.2.

developed in sub-section 6.3.3. Eventually, the results are presented in sub-section 6.3.4.

6.3.1 Transparent mode

In order to compensate for the undesired dynamic effects, the model described in 5.1.1 is used. The torque due to gravity, inertia and friction is then computed using an inverse model (calculated from the output angle and its first an second derivatives). It is then inverted and used as an input for the motor command.

6.3.2 Assistance mode

The proposed assistance mode uses a controller similar to the one proposed in (Lenzi et al., 2013). The hardware being different some details differ though.
The assistance controller works on top of the transparent mode. Thus, the friction, the gravity and the inertial effects are reduced as much as possible. As no force sensor is used in the HiBSO this ensures that the output torque closely matches the desired assistive torque.
The assistance is based on standard torque trajectories that were recorded on healthy subjects (Winter, 1991). These trajectories are presented on Fig. 6.3. Depending on the cadence,

different torque profiles need to be used. The torque is a cyclic function and it has the same period as the joint angular trajectory. From the flexion/extension angle, the cycle frequency can then be detected. The stride percentage can also be deduced from the angular position. The corresponding torque can thus be extracted. The cadence may take any value between approximately 80 and 120 *steps*/*min* and the stride percentage is also continuous between 0 and 100%. A function taking those two parameters was then fitted on Winter's data:

$$\Gamma_{assistance}(stride, cadence) = (A + B \cdot cadence + C \cdot cadence^2) \cdot \left(a_0 + \sum_{i=1}^{24} a_i \cdot cos(i\,\omega\ stride) + b_i \cdot sin(i\,\omega\ stride) \right) \tag{6.2}$$

A, B, C, a_i and b_i are the fitting parameters. ω is known as the period of the cycle is 100 (it is a percentage). It should be noted that the two parameters act independently on the torque function which corresponds quite accurately to what can be observe on Fig. 6.3. A 24^{th} order Fourier series was selected for the fitting of the cyclic part as relatively high frequency is required in order to fit the extension peak. A quadratic function is used to adapt this signal according to the walking cadence. The Fourier fit does not approximate perfectly the end of the cycle as the recorded torque profiles are not continuous at this point (i.e. the torque is not equal at 0 and 100%, see Fig. 6.3). This inconsistency in Winter's data may be due to the shock during the HS. Even though the Fourier fit is not perfect it was maintained as a periodic function reflects the intrinsic nature of the signal. This also guarantees smooth transitions of torque at HS.

The cadence and the stride percentage detection is based on the adaptive oscillators described in (Righetti et al., 2006). The input position signal is (quasi-)periodic and it can then be written as:

$$\theta_{input}(t) = \sum_{i=0}^{K} \alpha_{input\_i} \cdot sin(\phi_{input\_i}(t)) \tag{6.3}$$

$$\phi_{input\_i}(t) = i\,\omega_{input}\ t + \varphi_{input\_i} \tag{6.4}$$

where ω_{input} is the fundamental frequency and $i\,\omega_{input}$ the ones of the harmonics. α_{input_i} and φ_{input_i} are respectively the amplitudes the phases of the fundamental component and of its harmonics. It is therefore possible to learn these features with the estimator $\hat{\theta}(t)$ having the form of a Fourier series:

$$\hat{\theta}(t) = \sum_{i=0}^{J} \alpha_i \cdot sin(\phi_i) = \sum_{i=0}^{J} \alpha_i \cdot sin(i\,\omega\ t + \varphi_i) \tag{6.5}$$

where J is the number of harmonics kept in the estimator. As proposed in (Ronsse et al., 2013) the simplest system achieving the desired converging behavior is:

$$\dot{\phi}_i(t) = i\,\omega(t) + \nu_\phi \frac{F(t)}{\sum_j \alpha_j(t)} cos(\phi_i(t)) \tag{6.6}$$

$$\dot{\omega}(t) = \nu_\omega \frac{F(t)}{\sum_j \alpha_j(t)} cos(\phi_1(t)) \tag{6.7}$$

$$\dot{\alpha}_i(t) = \eta \cdot F(t) \cdot sin(\phi_i(t)) \tag{6.8}$$

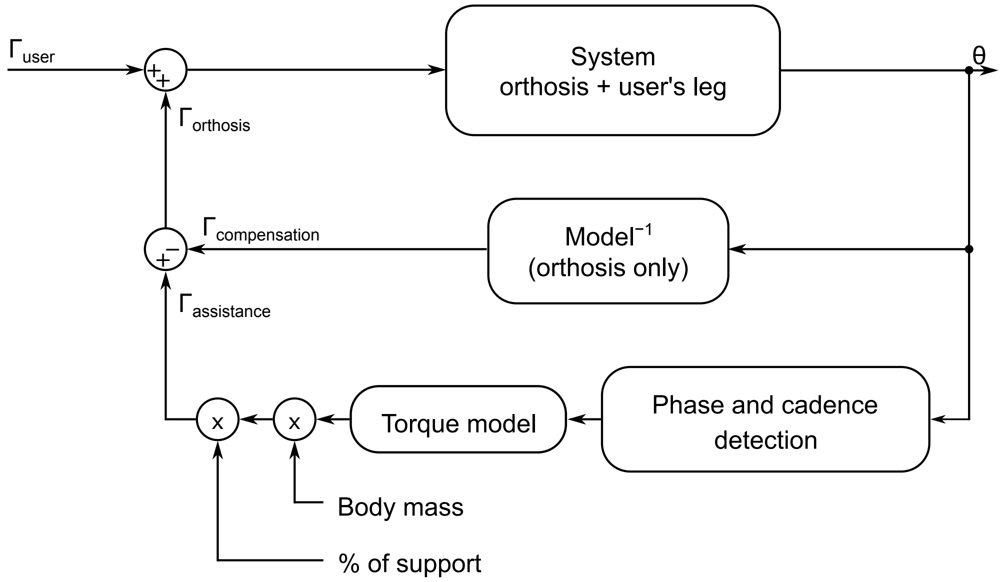


Figure 6.4 – Block diagram of the assistance controller. The assistance works in parallel with the compensation described in 6.3.1. The phase and the cadence are calculated from the orthosis flexion angle with the adaptive oscillators method. Then the model presented on Fig. 6.3 estimates the torque per *kg* body weight that the user needs to provide. This estimated torque is eventually rated by the user body weight and the percentage of support that needs to be provided.

with ν_ω, ν_ϕ and η being the learning gains. In that case $j \in [1;J]$. $F(t)$ is the error signal used for the teaching:

$$F(t) = \theta_{input}(t) - \hat{\theta}(t) \tag{6.9}$$

It can be seen from equations 6.6, 6.7 and 6.8 that the steady state is reached when $F(t)$ is equal to 0 (i.e. when $\hat{\theta}(t) = \theta_{input}(t)$).
The cadence in *steps/min* is then estimated directly from the the fundamental frequency of $\hat{\theta}$:

$$cadence = 120 \cdot \frac{\omega}{2 \cdot \pi} \tag{6.10}$$

The stride percentage is also estimated from $\hat{\theta}(t)$. To do so, the signal is evaluated on one complete period (calculated from ω) in order to detect a specific event. For the sake of simplicity and robustness, it was chosen to detect its global minimum which usually happens at approximately 54% of the stride. The current stride percentage is then deduced from the time difference with the global minimum. Note that due to the individual characteristics and soft tissues deformation this offset needs to be tuned on each subject. Concerning the latter, a delay usually appears as the mechanism always moves slightly relatively to the subject's leg. Fig. 6.4 shows the block diagram of the assistive controller.

Table 6.2 – Hip orthosis assistance experiment subjects' main characteristics.

Subject	Gender	Age (years)	Height (cm)	Weight (kg)
S1	Male	22	175	74
S2	Male	31	175	72
S3	Male	26	185	78
S4	Male	26	202	100

6.3.3 Method

Four healthy young subjects participated in this study. The main characteristics of the subjects are presented in Table 6.2. As the HiBSO interface was tailored on one of its developers, subjects with similar stature were selected. The subjective criterion of comfort was used as exclusion criterion. However it may be noted that a relatively large tolerance exists. Indeed subject S4 is 15% taller and 39% heavier than the person on whom the interface was tailored.
All recordings are performed on a treadmill at the imposed velocity of 1 m/s (3.6 km/h). Gait kinematics is recorded using the home made motion capture system presented in appendix A. First, a Free Walking (FW) session is recorded. Then, the subjects are asked to wear the HiBSO. Two sessions are recorded: one with the Transparent Mode (TM) controller and the second in Assistance Mode (AM) (10% of assistance). For these two sessions, an adaptation period of about 10 *min* is given to the subjects. The heart rate (HR) is also recorded in order to estimate the overall effort provided by the subject. This measurement is acquired during a separate session which is composed of 10 *min* of FW (FW-PRE) followed by 10 *min* with the HiBSO in TM and 10 *min* in AM. Eventually, 10 *min* of FW-POST are recorded so as to detect potential alterations due to fatigue.
For the AM condition, safety parameters were added on top of the controller. Three main conditions need therefore to be respected at all time. First, the gait cadence needs to be between 80 and 130 $steps/min$. Then, the absolute cycle offset between the two legs needs to be smaller than 10%. Eventually, the overall amplitude of the movement is required to be greater than 30°. To verify this, the sum $\sum_j \alpha_j$ is used. This prevents the assistance to be active when the subject is not walking. If any of the safety checks fails, the assistance is instantly turned off and the transparent controller (TM) is used. When the safety parameters are all respected, the assistance gradually increases so as to reach 10%. This ensures a smooth transition between TM and AM.

6.3.4 Results

Kinematics

In order to study the global influence of the device and its control strategies, the kinematic data of all the subjects are combined. The individual data are also verified so as to detect

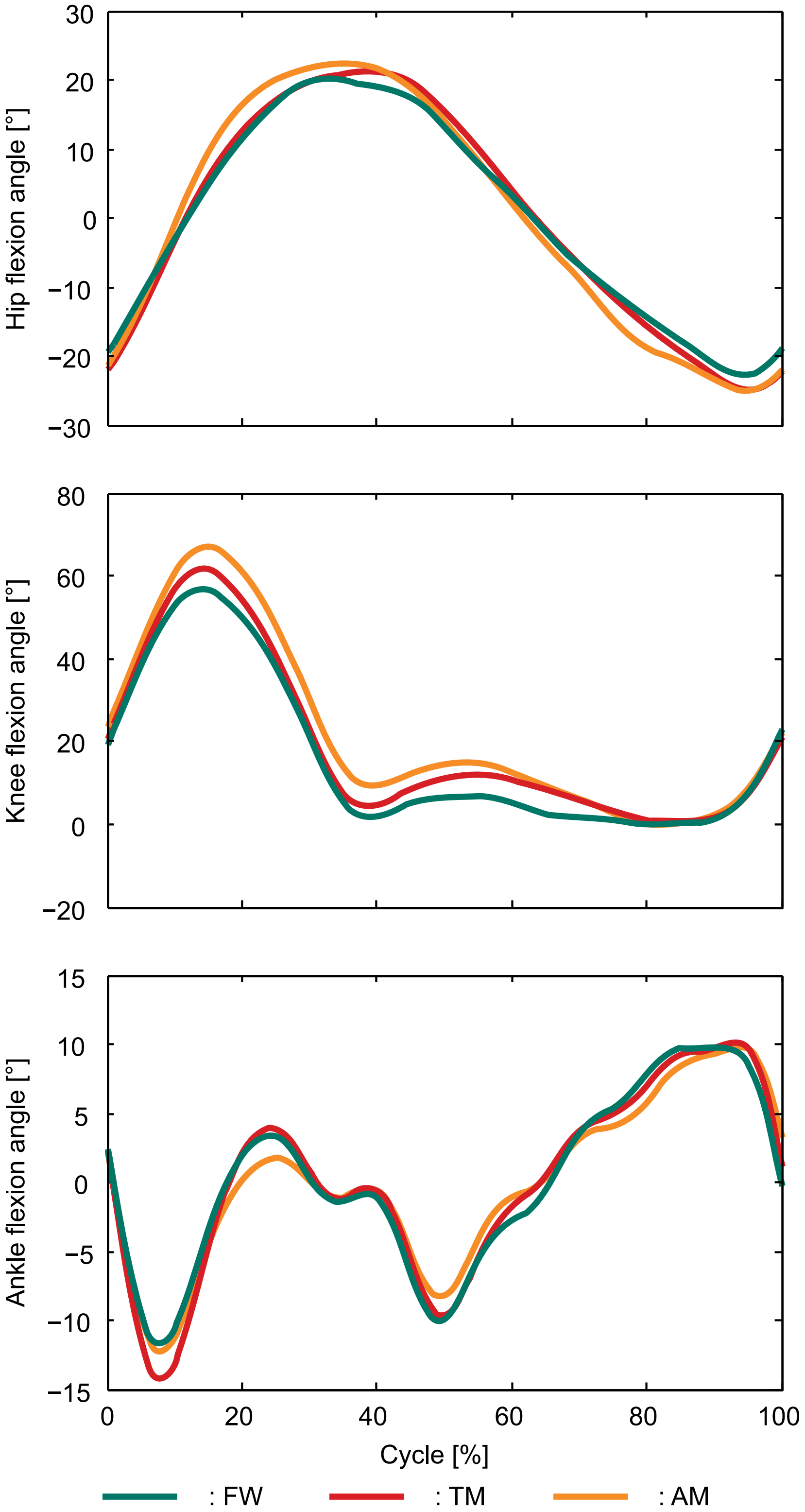


Figure 6.5 – Hip, knee and ankle trajectories in the three experimental conditions (combined data). The cycle starts at Toe Off (TO). The Free Walking (FW) condition is shown in turquoise, the Transparent Mode (TM) condition is in red and the yellow lines represent the Assistance Mode (AM).

any possible peculiar (or group) behaviors. The general tendency is therefore presented and distinctive features are mentioned when it applies. Fig. 6.5 presents an overview of the three leg articulations angles in the sagittal plane for the three different conditions (i.e FW, TM and AM).

As the device acts on the hips, an effect appears logically on this articulation. First, the RoM is larger in both conditions when walking with the HiBSO (augmentation of about 7% in TM, slightly larger in AM). The velocity and the accelerations during the swing phase (flexion movement) are higher in the AM condition than in any others. In TM, the tendency is not as clear. Indeed, even though the velocity is globally higher than in FW, the observation cannot be generalized to all the subjects. The extension (during stance) is also influenced with a slightly higher velocity in the AM condition, but to a lesser extent. The TM does not seem to affect the extension movement dramatically on any of the subjects.

Even though the knee is not assisted directly, a large influence can be observed at this joint. The HiBSO in TM induces an augmentation of about 6% of the knee RoM in average. Only one subject had a reduced RoM in that condition. In AM the augmentation is even higher and grows to about 16%. In that condition all subjects were influenced the same way.

The ankle is also influenced by the orthosis but it appears that the control mode plays a more important role in that case. In TM the RoM increases for most subject (+7% in average). In AM the RoM is not affected dramatically but the overall movement of the joint diminishes (compared to TM but also to FW). The velocity and the accelerations are then influenced by the two control modes. They are larger in TM and reduced in AM.

Other gait parameters such as the cadence or the step height do not exhibit a clear change tendency. Even though the variation may be relatively important for some subjects (cadence changes $\in [-8\%; 10\%]$ and step height changes $\in [-32\%; 27\%]$), not all the subjects reacted similarly and the average modification is therefore not representative. Considering this important variability, it may however be noted that different adaptation strategies are selected by the subjects.

Heart rate

The HR analysis clearly demonstrates that the walking task is more challenging in terms of energy consumption while wearing the HiBSO (it is reminded here that the energy consumption is highly correlated with the HR, see 1.3). The average HR (calculated over all the subjects) increases of approximately 10 $pulses/min$ between the FW conditions and the TM and AM conditions (see Fig. 6.6). The Student's t-test indicates that this increase is significant (p-value smaller than 5%). The difference between the TM and the AM (the two conditions involving the HiBSO) does not appear clearly. One subject only had a significantly different (lower) HR while wearing the orthosis in AM. However, his HR was still significantly higher in AM than in FW. The difference between the FW and the other conditions cannot be attributed to fatigue as FW before and after measurements with the HiBSO exhibit very similar values.

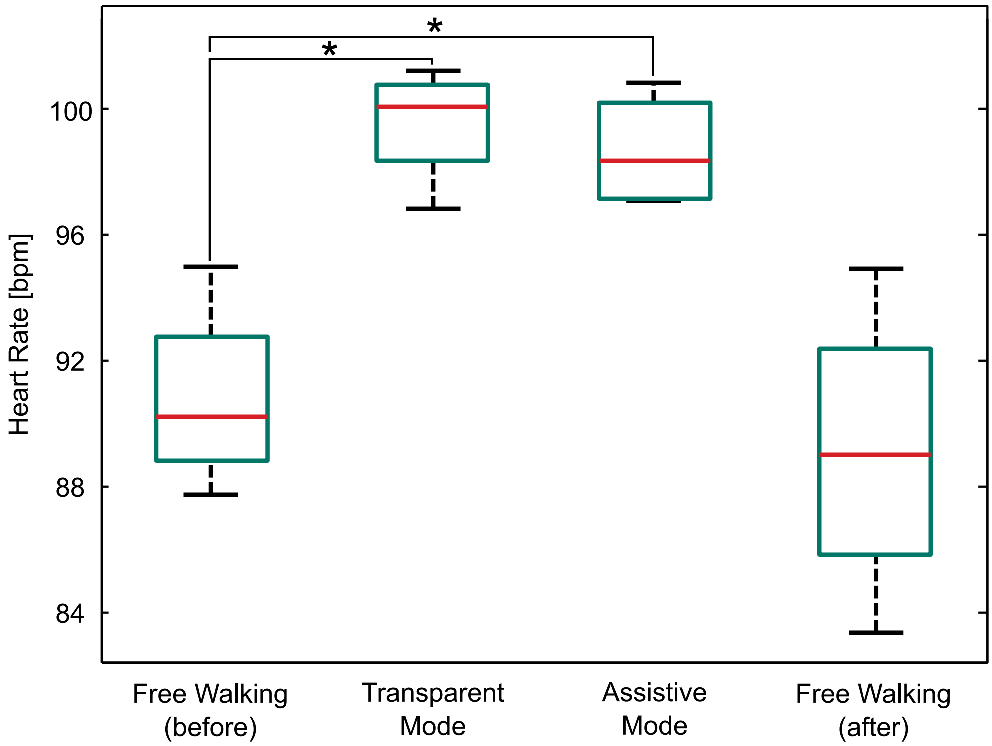


Figure 6.6 – Heart rate during the different experimental conditions (combined data). The FW condition is measured twice (before and after the two conditions with the orthosis) so as to detect possible fatigue effect. A significant difference can be observed between the FW conditions and the conditions with the orthosis.

6.3.5 Discussion

The recorded kinematic data demonstrate that the effect of a partial device goes beyond the joint it assists. Indeed the knee is the joint on which the device has the most influence in terms of RoM. The passive influence induces an augmentation which is further increased by the assistance controller. This can be explained by the fact that the acceleration of the hip flexion is higher during the swing phase, both in TM and AM. As the knee has a relatively low stiffness during this phase (Pfeifer et al., 2014), the motion of the thigh relatively to the shank is larger and the knee flexion is increased. Logically the subjects who increased the most their hip acceleration also had the largest increase in knee flexion.

The fact that the acceleration is larger during the swing phase in AM is not surprising as this movement is aided by the orthosis. However, the hip acceleration is also larger in TM during that phase, which implies that some subjects tend to generate an even greater torque at this joint than what would be required to compensate the added load. This overcompensation may indicate that these subjects aim to adopt a safe behavior where their hips are stiffened in order to reject the perturbations induced by the device. The velocity increase is smaller during the stance phase. This is probably due to the fact that the articulations are stiffer in stance and therefore less kinematic changes can be observed.

The kinematics of the ankle is also influenced by the device. In that case, the controller seems to play a major role as the TM and AM conditions have opposite effects on ankle velocity and acceleration when compared with FW. Even though this measurement does not represent a sufficient evidence on its own, it suggests that the ankle may benefit from this type of support. A study on the joints torques would then be required to confirm a possible reduction on the required ankle effort.

Attempts to reproduce the EMG measurements performed by Lenzi in (Lenzi et al., 2013) were also performed in order to manifest the differences in muscles activations (which are in direct link with joints torque and stiffness). Surface EMG signals of the Hamstrings (Semitendinosus), Quadriceps (Rectus Femoris), Gastrocnemius and Tibialis Anterior muscles were thus measured on several subjects during the study. This was done using a DataLINK system (Biometrics Ltd, *United Kingdom*). However, these measurements were unusable due to the to the low reliability of the surface EMG signal in amplitude. Indeed, the same condition exhibited drastically different amplitudes when measured at different points in time. A comparison between the different conditions would therefore be chancy and inaccurate. Moreover, external parameters (e.g placing the orthosis cuffs on top of the EMG electrodes attached to the thighs for two conditions only) may have an influence on the signal, which compromises its correct interpretation.

The HR measurements indicates that the subjects have to spend more energy to walk with the HiBSO than in the FW condition. This was also the case when assistance was provided. No significant influence of the assistance could be established in terms of metabolic cost even though one of the subject had a lower HR while being provided with assistance. This measurement being relatively uncertain, no conclusion can be made here. It may however be hypothesized that 10% of assistance with a device that adds an important extra load on

the wearer is not enough to assist gait properly. Two improvements are therefore required. First, the actuation needs to be able to provide more torque at the typical walking cadence. Second, the negative influence of the device needs to be addressed properly. The kinematics, the weight of the device and its interface with the body have to be considered carefully (see 6.2) but the actuation is also extremely important.
Preliminary HR tests have been conducted that compare the influence of the TM condition with the device without motor. These measurements demonstrated that the actuation unit is responsible for an important part of the metabolic cost increase (typically 50% on a typical subject). This suggests that mechanisms with more adapted actuation units could greatly improve the quality of assistive devices. Note that this also partially answers the question raised in 6.2. Indeed in the experiment performed with passive devices, the different tested conditions did not exhibit any clear influence on the kinematic patterns. No conclusion could however be made on the joints torque and stiffness nor on the metabolic cost. These preliminary tests demonstrates that the HiBSO has a significant influence on the metabolic cost. This implies that the subjects could reject the perturbations (no clear trajectory changes were reported) but more energy had to be spent.

7 Conclusions and outlook

Successful use of robotic orthoses for gait support has already been demonstrated in various cases. Indeed, this type of technology has acquired enough maturity to have actual positive impact on paralyzed patients. Commercial devices like the Ekso or the ReWalk have enabled these people to regain a type of locomotion that closely looks like normal walking (even though some differences still exist). The benefits of such orthotic devices is undeniable in this case as they restore part of a function that would otherwise be lost.

The improvements brought by these devices do not come without drawbacks. The additional load, the discomfort or the restrictions in the natural movements are only the first few examples of a long list. For the most severely impaired patients, these weaknesses appear marginal but most of the elderly do not belong to this category. In fact, in the case of people suffering from muscle weakness due to age, the benefits of commercially available orthotic devices would be completely outweighed by their negative aspects. Note that here the criticism is unfair as those devices are not intended for elderly people. However, a legitimate observation can be made that no existing device is designed so as to be effective with a person who is able to walk almost normally but needs help to gain confidence and autonomy[1].

The opinion that the negative aspects associated with exoskeletons are too detrimental for people being able to move by themselves is undoubtedly shared by the groups that develop soft exosuits (e.g Harvard Biodesign Lab or ETH Sensory-Motor Systems Lab). Evidently these devices do not suffer from the same type of inconveniences as the exoskeletons. However, our opinion is that this technology remains limited for assisting elderly people. Even though the level of constraint is very low, the tradeoff between this and the assisting capabilities is in our view not well balanced. Indeed, the working principle being to use natural level arms to provide extra torque to the articulations (Bartenbach et al., 2015b), relatively high forces are applied on the body and in particular at the joints[2]. In the frequent case of an elderly person suffering from Osteoarthritis (OA), such an extra load is clearly not desirable. The large number of military applications being published creates the impression that mainly healthy

[1] Rare exceptions exist as for example the Honda stride management assist. Scientific demonstrations of the effectiveness of these devices are however still missing.

[2] The more conventional orthoses usually have longer lever arms and therefore the reaction forces at the joints are supposedly lower than that induced by the exosuits or even the muscles.

people could benefit from such devices[3].

7.1 Main contributions

In this thesis, the objective was to develop orthotic solutions for elderly people. The approach of using rigid structures acting in parallel with the body was adopted. In order to fit the specific needs of the population of interest, it is important to identify the main sources of perturbations induced by exoskeletons. Since, for seniors, a moderate but non negligible level of assistance is required, assumptions were made to optimize the tradeoff between invasiveness and proportion of support. Logically the presented developments are an attempt to reduce the perturbations at the most while preserving the possibility to efficiently support the weak muscles. While devices for handicapped people (e.g. Spinal Cord Injury (SCI) patients) need to lock certain Degrees of Freedom (DoFs) for stability, the prototypes that we present here tend to preserve all the possible movements without restraining their Range of Motion (RoM).

7.1.1 Partial assistance

One of the key concepts that is investigated in this thesis is the effect of partial assistance. By limiting the amount of assisted DoFs, the number of constraints is naturally reduced when compared with devices that support most of the DoFs of the lower limbs (e.g. REX). Going one step further in this direction, the number of assisted articulations can be reduced as well. A device acting on the hip only is likely to be less invasive than devices acting on all the leg's articulations (e.g. ReWalk, Ekso) even if the number of assisted DoFs is limited. Nevertheless, as a partial device cannot be self-supported, a particular attention should be paid to limit its weight.
A study has therefore been conducted to show the effect of a wearable device acting on the hip only. This study was performed with young and healthy subjects. In addition to the weight, effects such as the kinematic constraints or the interface with the body were investigated. Note that these aspects are common to all orthotic devices, partial or not. In the study, two different orthotic devices and a total of six different conditions were used to better highlight the impact of every single factor (namely the weight, the kinematic constraints and the interface with the body). With the task that the subjects were asked to perform (i.e. walking on a treadmill at their self-selected velocity), none of the conditions had a significant effect on the kinematic patterns of gait. While this observation may appear surprising, it is a valid indicator that the tested devices did not perturb the subjects in a dramatic manner when they were walking on a treadmill (at least at the kinematic patterns level).
The second hypothesis that needs to be tested in the context of partial devices is the influence

[3]It is also communicated that these devices will be used with elderly people but no demonstration has been made yet.

of an incomplete assistance. The assumption that was made is that a device that assists the hip in the sagittal plane should have a positive impact on frail people as it supports the compensation strategy that this population usually adopts. The weaknesses at the lower limb appear more severely at the ankle joint and the hip is consequently more solicited. We decided to assist the hip instead of the ankle so as to ease the implementation of the compensation strategies instead of trying to restore a nominal gait[4].
A study investigating the effects of a moderate assistance (10% of the nominal torque) at the hip level was thus conducted on healthy subjects. In this experiment, the assistance simply follows standard torque curves from the literature. This torque profile is then synchronized with the subject's strides. The results of this study demonstrated that the assistance has a significant effect on the ankle trajectory. It may then be hypothesized that the ankle could benefit from this type of assistance. However, as the joint torques were not measured, no confident conclusion can be made about any actual strength changes.
Another result of this study concerns the heart rate (HR). The assistive device has a notable negative impact on this measurement. This observation is valid when no assistance is provided but also when 10% of support is transferred to the hip. It therefore appears that the subjects were not able to substantially benefit from the assistance (at least from the metabolic cost point of view). Multiple rationales could explain this. First, the level of assistance being relatively low, its effect may be too little to be perceivable. Second, the period of adaptation may have been too short for the subjects to adapt and to be confident in the device effect. Third, the assistance may be perceived as a perturbation by the user and therefore be rejected instead of being exploited. The human motor control being more complex than simply applying torques at the articulations to produce the desired trajectories, the users may not be extremely confident in a device that simply adds torques at their hips.
Further investigation is therefore required so as to address these issues. As only 10% of assistance was provided in our study, a more vigorous support is evidently to be tested. Other research directions are equally important and this is further discussed in 7.2.

7.1.2 Innovative transmissions

In the proposed developments, a peculiar effort was made on the actuation which is often overlooked in orthotic devices. Indeed, standard reduction gear trains are, in our opinion, poorly adapted to the specific needs of assistance.
In this field, the first improvement that we propose is a transmission that varies as a function of the position. As the needs in terms of peak flexion/extension torque tend to be correlated with the position, such a transmission ratio is helpful to increase the number of activities that can be assisted. Moreover, the dynamic requirements being more strict in the walking RoM, an adaptation of the capabilities in this area is highly recommended. This combination of

[4] This choice is motivated by the fact that elderly people are usually subject to a combination of conditions and their gait modifications are adapted to their abilities. Assisting their compensation strategies may therefore be less destabilizing than restoring the capabilities of the most affected joints.

specifications perfectly fits with what can offer a transmission whose reduction depends on the position. An implementation of this concept was developed and a functional prototype, the Hip Ball-Screw Orthosis (HiBSO), was used for testing the above-mentioned assistance strategy on healthy subjects.
A second prototype was build based on another transmission concept. The Double Differential driven Orthosis (DDO) uses a transmission featuring two clutches and an inversion mechanism, which makes the decoupling of the actuator and the end-effector possible (i.e. the part attached with the user's leg). The inversion mechanism is used to avoid reversing the direction of rotation of the actuation unit. This enables to save energy by circumventing the undesirable effects of the motor inertia. As mentioned in our studies , the sub-obtimal transparency induced by the actuation unit is responsible for a significant proportion of the metabolic cost increase when wearing a robotic orthosis (see 6.3). Improvements in terms of transparency are therefore highly recommended and such developments typically address this issue. The second advantage of our mechanism when compared with standard reduction gear trains is that extremely demanding acceleration profiles can be followed without requiring much energy as the actuator never needs to vary its velocity (see 5.1.3).
A third development based on a concept working on an inversion mechanism and a spring was introduced but no prototype was built. The idea behind this concept is that energy can be harvested when the required power is negative. In some of the gait phases, in order to support the articulation, energy needs to be subtracted from it. This happens during the phases where the articulation exhibits negative power. With an adequate mechanism, the energy that is normally dissipated by the joint can be transferred to an elastic element so as to be temporary stored. It can then be used when needed, i.e. when positive mechanical work is required (which corresponds to the gait phases where joint power is positive).
During walking, dissipative phases exist for all three lower limb articulations but this is particularly valid for the knee. Indeed this articulation exhibits more negative than positive mechanical work on one complete walking cycle. In other terms its Root Mean Square (RMS) power is negative. The idealistic concept is therefore to store enough energy during the phases were energy can be subtracted from the joint in order to return it during the positive work phases. In the case of the knee, extra energy should even be dissipated, for instance in a brake. Consequently, such a device could theoretically work without motor. Similarly to an automatic mechanical watch, the energy would be stored in a spring that would be loaded by the wearers themselves. The only elements requiring electrical energy would be semi passive components to control the energy flow. The analogy with the watch stops here as, to display the time, the energy needs to flow in the mechanism in a predefined manner.
In the proposed concept, the spring can be coupled or uncoupled by means of a mechanism similar to the DDO. Torque in both directions can thus be provided. The regulation of the output torque is done by using the two control brakes (and possibly a third brake to preserve the energy stored in the spring). In our design, a motor was also used in order to load the spring (in case the energy harvested from the assisted joint does not suffice for the positive power phases).
The required complexity for an actual implementation appeared to be too high with respect

to the expected benefits. This concept however highlights several important points. First, the energy that needs to be stored is not extremely high and elastic elements of a decent size can be used for that purpose. Second, even though the inversion mechanism appears to be insufficient on its own to efficiently control the energy flow, energy can be provided and removed from the assisted joint and in some (regrettably too rare) cases transferred to the spring.

The logical observation is that a single transmission ratio is not sufficient and using a clutch for compensating for this lack is evidently not optimal. A gear box would then be helpful as described in (Beachley and Frank, 1979). An even better component would be an Infinite Variable Transmission (IVT). With such a system, the clutches could even be avoided[5].

7.2 Outlook and future research directions

7.2.1 Constraints caused by orthotic devices

The negative aspects induced by robotic orthoses have often been neglected even though they may dramatically affect the overall usefulness of a device. Identifying them in order to modify the exoskeletons design may have vast implications for all categories of devices (i.e. rehabilitation, mobilization, assistance and performance augmentation). Within the framework of a collaboration with the Sensory-Motor Systems Lab from Professor Riener, we have made a first step towards a better understanding of the mechanisms responsible for affecting the wearer of an orthosis.

The undeniable limitation of our study is that only a very simple task was tested: walking on a treadmill. Even though the kinematic patterns were not significantly affected, any conclusion stating that hip orthoses do not affect gait should be moderated for two main reasons. First, neither the metabolic cost nor the joints torques were measured. The muscle co-contraction may also be modified. Our assumption is that all these factors are affected as a perturbation exists and appears to be rejected. Concerning the metabolic cost, preliminary tests have been conducted that show a significant increase of the HR[6] when wearing the HiBSO (without its actuation unit). Second, the evaluated task takes place mainly in the sagittal plane. The requirements in terms of hip adduction/abduction and internal/external rotation in amplitude are therefore limited. Furthermore the trunk does not need to move considerably during this simple task. Different activities such as standing up from a sitting position, climbing/descending stairs or even walking on a non-straight trajectory would certainly generate different results.

Other complete and partial devices for the knee and the ankle should also be evaluated. Research platforms such as the one presented in (Bartenbach et al., 2015a) are valuable tools for isolating the effects of every possible source of constraint. Lighter structures, novel kinematic designs or improved body interfaces may also be required so as to limit the perceived invasive-

[5] It is reminded here that a transmission ratio which divides the velocity by zero corresponds to locking the input and uncoupling the output.

[6] The HR is correlated with the metabolic cost in absence of cardiac disease.

ness of orthotic devices.

7.2.2 Assistance

The seamless integration of an assistive device into the motor control of its user is probably one of the major challenges that remains to be addressed in assistive devices. In the case of paralyzed people, only high level commands can be generated by the users. The challenge is therefore to detect their intentions as intuitively as possible. In the case of less severely impaired patients (in our case, elderly with reduced vigor), the difficulty is not exactly to detect the intentions but to work in collaboration with the weak joints. Therefore only relatively low-level control needs to be implemented. Note that higher-level signals could be used as well but they are usually more difficult to interpret. Moreover their signal extraction tends to be more invasive (e.g. electrodes are required for Electromyography (EMG), hat with electrodes are required for Electroencephalography (EEG), etc.). As our target users are still able to act at the same level as the device, they may also react to it. A bi-directional adaptation is then required.

The controller that we implemented takes the position of the device as only input. We assumed that this position relatively closely matches flexion/extension of the hip. During walking, the assistive torque is synchronized with the position trajectory. While this controller had a demonstrated impact on the hip-ankle coordination, we could not highlight a positive impact on the metabolic cost. One of the possible explanations for this is that a controller mimicking the torque is not well accepted by the users as it does not act similarly to the intact human motor control. We therefore hypothesize that it would be useful to investigate the effect of a controller that reproduces not only the torques of the joint but also its complete impedance behavior. This evidently implies that these behaviors need to be well mastered. Devices for measuring the impedance of biological joints are therefore required (Tucker et al., 2014). Moreover, it is important to consider that the impedance is influenced by the task that is being performed and even by the expected conditions. For example, it is well known that walking on a slippery surface modifies the muscular co-contraction and therefore the impedance at the different joints.

Working on a controller that reproduces a variable impedance and that takes into account the conditions (e.g. slippery or uneven terrain) could also be useful in order to anticipate tripping or slipping. The safe behavior adopted by people walking on challenging surfaces could thus be managed by the device; the users could then possibly save energy and gain in autonomy and confidence.

Fall recovery is an even greater challenge. In the same manner as humans do (see 1.4.3), three main conditions have to be fulfilled to address this problem. First the risk of fall needs to be detected. Appropriate sensors and algorithms are thus required. This point represents a challenge on its own. Then an appropriate response is required. For a device, it means that a safe strategy has to be known or possibly developed online. Note that here again, this strategy needs to take into account the actions of the users and work in coordination with them. Third, this response needs to be implemented quickly enough to prevent any too hazardous situation.

7.2.3 Actuation

Fully agreeing with (Dresscher et al., 2015), we suggest here that the assistive orthoses developers should investigate the feasibility of compact IVTs that are able to amplify a few times the input torque as well as to reverse it. Elastic elements could then be used to harvest energy during the phases of gait where negative power is required. In that case, large transmission ratios would not be necessary as elastic element can usually provide much higher torques (or linear forces) than electric motor (without gear box) of the same volume. Moreover, the large amount of energy that is usually spent on the motor dynamics itself could be greatly reduced as proposed with the DDO mechanism.

7.2.4 Clinical evaluation on elderly subjects

In order to make sure that the partial assistance at the hip is indeed beneficial for the target population, clinical tests with elderly patients should be conducted. Even in the idealistic case where tests on healthy subjects would demonstrate excellent results (e.g. reduction of the solicitation at the different joints for various activities, reduction of the energy consumption during locomotive task, etc.), the seniors may react differently. In this work we hypothesized that partial assistance may facilitate the implementation of compensation strategies. Evidences tending to show that the effects of the provided assistance spans more than the supported joint exist. Nevertheless these results were obtained on young and healthy subjects. The muscles of elderly people tend to be weaker than that of younger adults. Logically the assistance should then be helpful. However, as their motor control and their coordination may also be affected, it would not be safe to guarantee that the effect would be the same on this population. The integration of the device could then be performed less effectively.
In addition to the evaluation at a local level (i.e. joint kinematics and torques, muscle activation, standard gait parameters, etc.), standard evaluation tests such as the Performance Oriented Mobility Assessment (POMA) must also be conducted. Indeed, the most important parameter for the elderly is the functional performance in Activity of Daily Living (ADL) and not necessarily an improvement of one particular factor. Therefore only pragmatic tests can assess actual benefits of an assistive device.

In conclusion, the developments and the studies presented in this work are a first step towards simple, light and effective assistive devices for elderly. The specifications that were established based on literature review and discussions with gerontology and rehabilitation specialists appear to be reasonable as the proposed prototypes showed a reasonable invasiveness. The benefits of partial assistance could not yet be unequivocally proven but preliminary evidences exist that such a support could be helpful for the elderly suffering from muscle weaknesses.

Moreover, innovative transmissions for robotic orthoses were also proposed. These new concepts were developed considering the elderly needs but they could also be beneficial for other human-machine interfaces where bi-directionality is essential.

A Motion tracking system

In this appendix, a "home made" motion tracking system is presented (Nguyen et al., 2012). It was developed to facilitate gait kinematics evaluation. The system is very compact as it uses a single tracking unit and seven active markers that need to be placed on the different body segments. In the following section the tracking device functioning is explained. Then the methods used for the extraction of the important gait parameters are developed. Another feature regarding skeleton reconstruction is explained in A.3. Eventually some test recordings are presented.

A.1 Tracking setup

The tracking is based on a commercial solution: the AccuTrack250® (Atracsys, *Switzerland*). This device uses active wireless markers and a single tracking unit composed of three linear cameras. The tracking unit size is 290 *mm* x 70 *mm* x 95 *mm* and its weight is 1 *kg*, which makes it easy to transport. Moreover, it can easily be installed on any standard camera tripod. Its maximal acquisition frequency is 4112 *Hz*. The measurement precision is 0.14 *mm* RMS up to 2 *m* and 0.21 *mm* RMS up to 2.5 *m*.
The device workspace is presented on Fig. A.1. As this workspace is limited it is important for the tracking unit to stay relatively close and oriented towards the tracked objects. Therefore, most of the time the motion capture is performed on a treadmill. However, a version using a mobile frame has also been implemented to study overground walking (Ortlieb et al., 2014).
The main advantage of using active markers is that they have a unique Identifier (ID). In the case where a marker is lost (due to an occlusion for instance) it is instantly recognized when it is detected again. This simplifies greatly the body parts recognition algorithm. Occlusion may happen when an object passes in between the tracking unit and a marker, if a marker is out of the workspace or if the orientation of the marker is too important. In the latter case, the marker hides its own LEDs. It therefore depends on its geometry but also on its position as the vision orientation is not homogeneous in the entire workspace.
On each marker, four IR LEDs are positioned (see Fig. A.2), which enables to define a rigid body (a calibration is made once for each marker). With only three tracked points, the markers

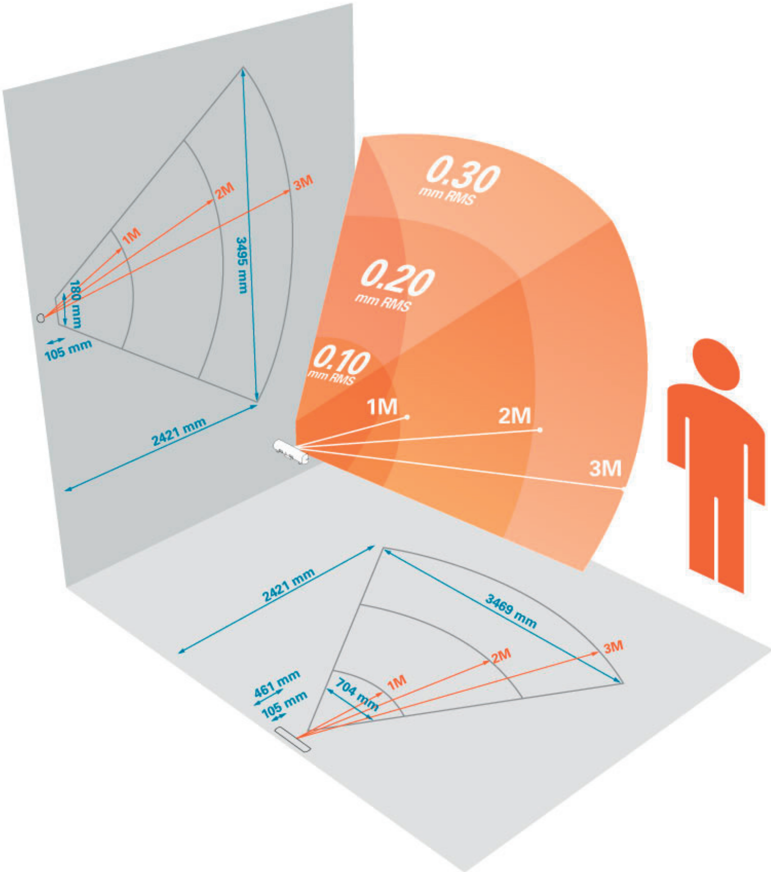


Figure A.1 – AccuTrack250 workspace. Image from http://www.atracsys.com/.

Figure A.2 – Atracsys wireless marker. Image from http://www.atracsys.com/. Four Infrared (IR) Light Emitting Diodes (LEDs) are used to enable the tracking of the position and the orientation of the markers.

could be localized in position (three-dimensional (3D) coordinates) and orientation (rotation matrix). With these markers geometry, the angular precision is about 0.4 °. The fourth LED is redundant and thus used to estimate the measurement error whose main source is the marker movements. Indeed as the LEDs are tracked sequentially, the movements induce inaccuracies in the measurement (the different points on the marker are not measured at the exact same time). Another consequence of the sequential measurement is that the overall frequency is divided by the number of tracked points. In case where seven markers with four LEDs are used, the frequency drops to about 150 *Hz*.
In order to study the lower limb kinematics, markers have to placed on the body segments that are adjacent to the articulations. To track the movements of the hips, the knees and the ankles, markers need to be placed on the feet, the shanks, the thighs and on the pelvis. In total, seven markers are required (see Fig. A.3).

A.2 Articulation angles calculation

To be able to calculate the articulation angles, a reference posture (where all the joints have a zero angle) has to be measured first. This guarantees that no offset is induced by the markers placement. At any point in time, the rotation matrices of the markers therefore need to be adjusted by the reference offset matrices. To calculate the angles of a joint, the two adjusted rotation matrices of the body segments adjacent to the considered joint are used. The rotation matrix describing the joint R_{joint} is then simply the matricidal product of the moving segment rotation matrix $R_{moving\_segment}$ and the transposed of the reference segment rotation matrix $R_{reference\_segment}$:

$$R_{joint} = R_{moving\_segment} \cdot R^T_{reference\_segment} \tag{A.1}$$

The reference segment is the pelvis for the hips, the thigh for the knees and the shank for the ankles. From the joint rotation matrix, the Euler angles can be calculated easily. Note that here the Euler angles are preferred to the projected angles describe in 1.3.1 as they are less ambiguous when several rotations are combined. Eventually, minor sign adjustments need to be done in order to be coherent with the literature (see 1.3.2).
The beginning of the cycle is defined by the Right Toe Off (RTO). This event was chosen as it can be detected by a simple condition on the position of the right foot marker. Indeed, when walking on a treadmill the position of the foot reaches a minimum when the leg starts the swing phase (see Fig. A.4). Thanks to this event detection, the cycles can be normalized. The mean trajectory (and a confidence interval) can be estimated and all the standard gait parameters described in 1.3.1 can be calculated.

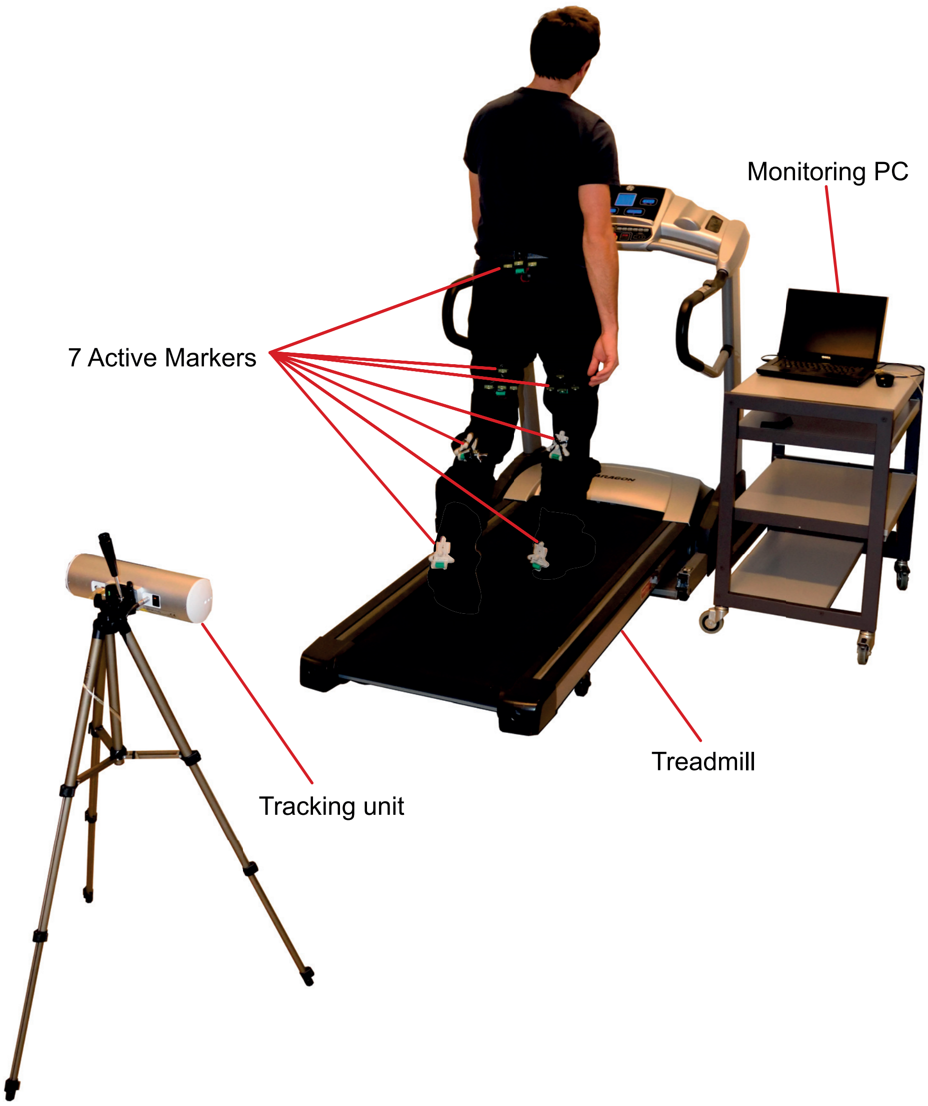


Figure A.3 – Motion capture setup. The tracking unit is placed behind the subject who is walking on a treadmill. Seven active markers are positioned on the different lower limb parts (pelvis, thighs, shanks, feet).

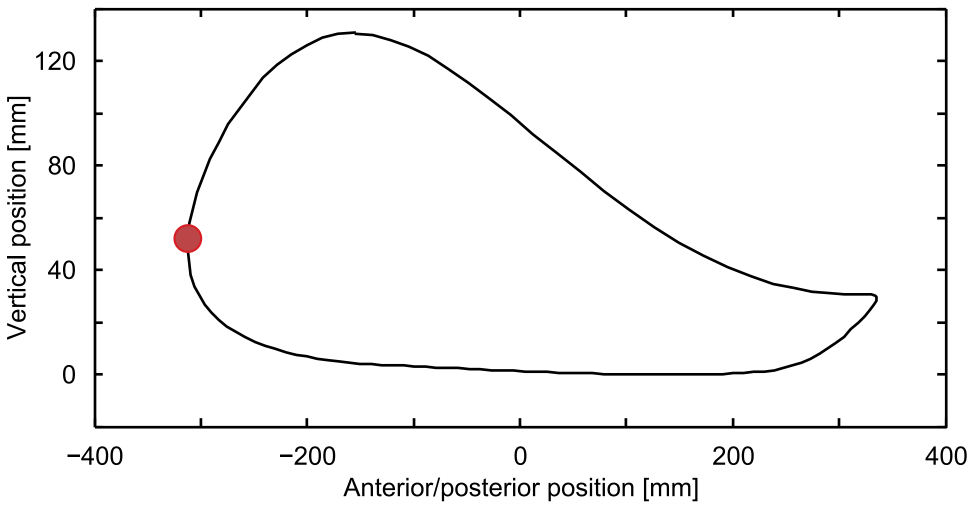


Figure A.4 – Trajectory of the foot (in that case malleolus) when walking on a treadmill. The minimum antero/posterior position is reached when the leg starts the swing phase (red circle).

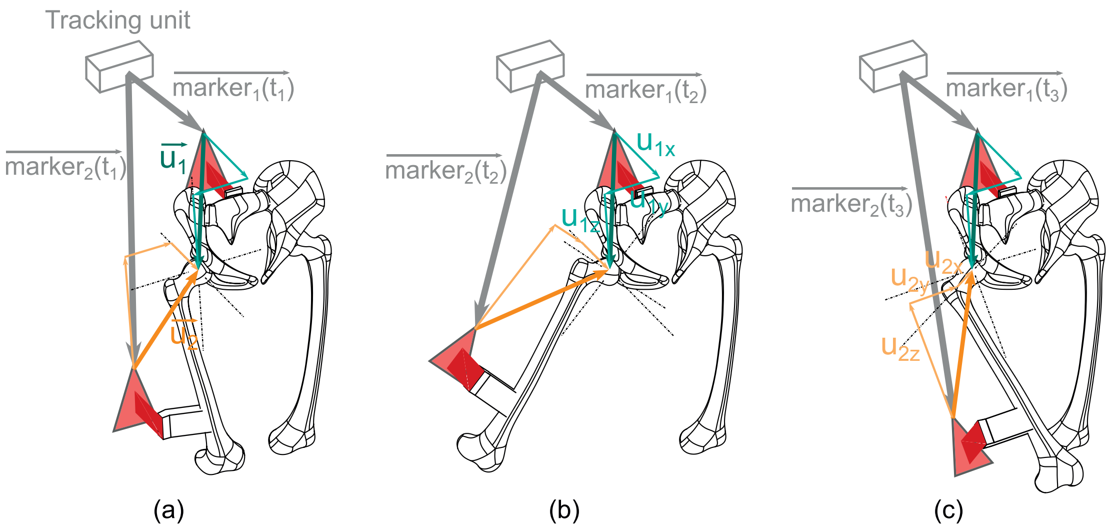


Figure A.5 – Position of the articulation in the referentials of two markers. The markers are shown in red. Sub-figures (a), (b) and (c) show different postures at different times t_1, t_2 and t_3.The two vectors $\overrightarrow{u_1}$ and $\overrightarrow{u_2}$ do not change over time in their respective referential but $\overrightarrow{marker_1}$ and $\overrightarrow{marker_2}$ do (tracking unit referential). From several acquisitions of the two vectors $\overrightarrow{marker_1}$ and $\overrightarrow{marker_2}$ and the rotation matrices R_{marker_1} and R_{marker_2}, $\overrightarrow{u_1}$ and $\overrightarrow{u_2}$ can be estimated.

A.3 Skeleton reconstruction

In order to better visualize or to play back a recorded session, it may be useful to reconstruct the body segments. To do so, the positions of the articulations are estimated. It is assumed that the markers do not move relatively to the positions of their adjacent articulations (rigid links assumption). Therefore, the position of an articulation in the referential of its neighboring markers is fixed. When the articulation DoFs move, the two corresponding markers move relatively to each other observing this constraint. This can be formulated as follows:

$$\underbrace{\overrightarrow{marker_2(t=t_i)} - \overrightarrow{marker_1(t=t_i)}}_{B_i} = R_{marker_1}(t=t_i)\cdot\overrightarrow{u_1} - R_{marker_2}(t=t_i)\cdot\overrightarrow{u_2} \tag{A.2}$$

$$= \underbrace{\begin{pmatrix} R_{marker_1}(t=t_i) & -R_{marker_2}(t=t_i)\end{pmatrix}}_{A_i}\cdot\underbrace{\begin{pmatrix}\overrightarrow{u_1}\\ \overrightarrow{u_2}\end{pmatrix}}_{\overrightarrow{X}} \tag{A.3}$$

where $t = t_i$ indicates the time when the acquisition is recorded, $\overrightarrow{marker_1}$ and $\overrightarrow{marker_2}$ are the positions of the two markers in the tracking unit referential. R_{marker_1} and R_{marker_2} are the rotation matrices of the two markers. $\overrightarrow{u_1}$ is the position of the articulation in the first marker referential and $\overrightarrow{u_2}$ is its position in the second marker referential (see Fig. A.5). A_i and B_i are the matrices that will be used for solving the system. Their dimensions are respectively 3×6 and 3×1.
By taking several acquisitions with different articulation angles, $\overrightarrow{u_1}$ and $\overrightarrow{u_2}$ can be estimated precisely. In fact, each acquisition i generates new matrices A_i and B_i. In order to make sure that this calculation is correct for various articulation angles a system of equations based on n acquisitions is built:

$$A = \begin{pmatrix} A_1 \\ \cdots \\ A_i \\ \cdots \\ A_n \end{pmatrix} \qquad B = \begin{pmatrix} B_1 \\ \cdots \\ B_i \\ \cdots \\ B_n \end{pmatrix} \tag{A.4}$$

This system is solved by using the method of least squares. A pseudo inverse matrix is then built. Eventually, the solution is given by:

$$\overrightarrow{X} = A^T\cdot(A\cdot A^T)^{-1}\cdot B \tag{A.5}$$

The dimensions of A and B are $3\cdot n\times 6$ and $3\cdot n\times 1$.
Once $\overrightarrow{u_1}$ and $\overrightarrow{u_2}$ are calculated, the error on the articulation position can be estimated. This error is estimated by the difference between the positions calculated from two different

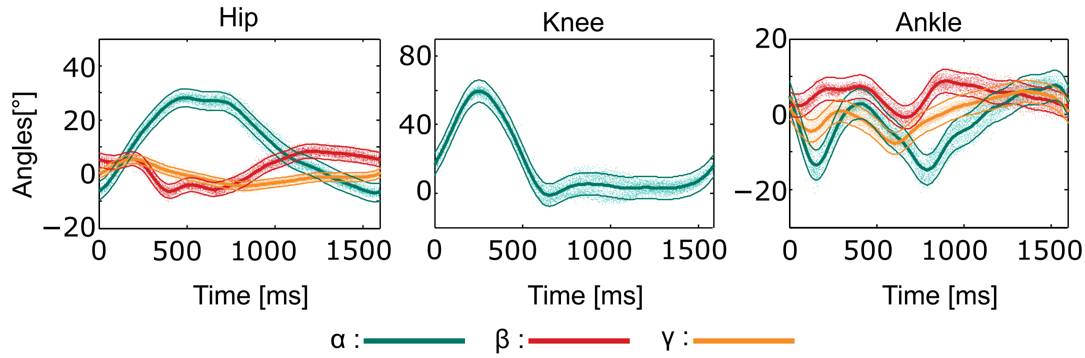


Figure A.6 – Acquisition of hip, knee and ankle Euler angles during walking. Cycles start and end at Toe Off (TO). Turquoise lines are the α Euler angles which correspond respectively to the flexion/extension of the hip, the flexion/extension of the knee and the plantar/dorsi flexion of the ankle. Red lines represent the β Euler angles and yellow lines are the γ Euler angles. The dots represent the measurements, the thick and the thin lines are respectively the 7^{th} order Fourier series fit of the data and the 95% confidence intervals.

markers:

$$\overrightarrow{{}_1joint} = \overrightarrow{marker_1} + R_{marker_1} \cdot \overrightarrow{u_1} \tag{A.6}$$

$$\overrightarrow{{}_2joint} = \overrightarrow{marker_2} + R_{marker_2} \cdot \overrightarrow{u_2} \tag{A.7}$$

$$error = \left\| \overrightarrow{{}_1joint} - \overrightarrow{{}_2joint} \right\| \tag{A.8}$$

where $\overrightarrow{{}_1joint}$ and $\overrightarrow{{}_2joint}$ are the positions of the joint calculated from one marker or the other. This method is applied to the six considered articulations of the legs. To do so, the subject is asked to perform random movements using all the possible lower limb DoFs. Without surprise, the method works very well for the hips and the ankles as they have multiple DoFs. Therefore, more than one axis of rotation can be found. Their intersection corresponds then to the position of the articulation. Interestingly, the method works also very well for the knees. Logically only one axis should be found and therefore the position of the articulation should be estimated randomly along this line. However this is not the case and the articulation is positioned at its correct location. This is mainly due to the rotational movements of the shank about the longitudinal axis.

A.4 Results

Fig. A.6 shows the typical curves that can be obtained with the motion capture system based on the Atracsys measurement solution. As the cycles are normalized, additional parameters such as the average cadence and its standard deviation are calculated. The angular velocities and accelerations are calculated based on the 7^{th} order Fourier series fits.
The 3D reconstruction of the skeleton is a valuable tool to playback a recorded session. It is also useful during the acquisition to detect any recording problem (e.g. incorrect position of a marker). Moreover, as the joint position error is very sensitive to the markers orientation, it

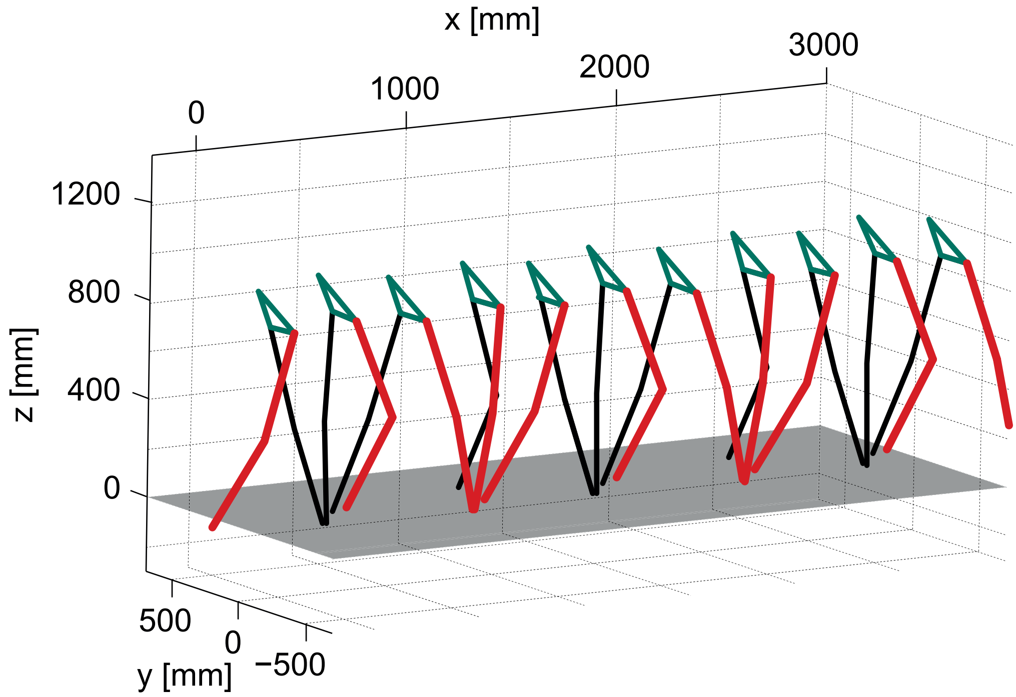


Figure A.7 – 3D reconstruction of the subject's skeleton. The different segments link the estimated positions of the articulations. The feet are not displayed as the position of the toes is not precisely known. The third point at the pelvis level is the position of the pelvis marker (only used to facilitate the representation of the lower limbs).

enables to detect if the markers have moved too much during the recording. An error of 10 *mm* on the articulation position is therefore acceptable as it represents typically a couple of ° on the marker orientation. Moreover, the error may also come from position errors which do not influence the angle measurement.

B Impact of Ankle Locking on Gait

This appendix is an adaptation of (Olivier et al., 2015) and presents a study on the locking of the ankles (and feet) and its influence of the kinematics of gait. It is motivated by the fact that most of the commercially available exoskeletons for paraplegics have four active DoFs (see 2.1), which enables to mobilize the hips and the knees in the sagittal plane. The exoskeleton ankle joints are usually passive and their motions are therefore limited if not completely locked. This strategy was adopted by several development groups as it appears that the movement of the ankle may remain unactuated without compromising the user's safety (Veneman et al., 2007). However, the nominal trajectories of the two other articulations may lead to a sub-optimal gait and they should be adapted.

In order to generate trajectories for such exoskeletons different strategies may be adopted. Some of them calculate the trajectories online (e.g. the Zero Moment Point (ZMP) (Wang et al., 2013)). Usually these control strategies are based on stability concepts. The main drawback is that the gait patterns do not appear natural. Another very common technique uses purely predetermined paths which come from simulations or simply by capturing the gait trajectories of healthy users wearing the exoskeleton. The latter is attracting as it easily provides natural human movements. However, it requires that the system is already built and that the user is able to back-drive the actuation if it cannot be removed.

In order to generate the trajectories before having the exoskeleton built, it is proposed to study how a simple ankle locking acts on the hip and knee joint trajectories. Even though the locking of the adduction/abduction of the hip is not provided with such a technique (as would be with the actual exoskeleton), it appears as a reasonable trade-off for estimating the final optimal trajectories. In this study, ski boots are used as they have a high stiffness and they also lock efficiently the movement of the feet (stiff soles).

Note that while this study does not address the same concern as the rest of the thesis, some aspects are in direct link with it. Indeed, even though the exoskeleton designs that this study addresses are made for paralyzed patients, the negative impact of a kinematic design on gait is strongly related with the main matter presented in this work.

Table B.1 – Ankle locking experiment subjects' main characteristics.

Subject	Gender	Age (years)	Height (cm)	Weight (kg)
S1	Male	31	175	71
S2	Male	23	183	69
S3	Male	26	185	78
S4	Female	32	169	62
S5	Male	26	165	45
S6	Male	26	189	78
S7	Male	22	175	74

B.1 Methods

B.1.1 Task

In order to study the influence of ankle and foot locking on gait, kinematic data are acquired when walking with ski boots. The acquisition is performed with the system described in appendix A. These data are compared with baseline (BL) recordings in order to detect the induced modifications. The session therefore starts with these BL recordings. This enables the subjects to familiarize with the experimental setup. For the sake of simplicity, the experiment is performed on a treadmill even though overground recordings could have been done with the system presented in (Ortlieb et al., 2014). Note that the treadmill recording also enables to record more strides in the lab environment. The subjects have therefore more time to get used to the setup in general but also to the locking condition. While this represents a limitation (mainly because the treadmill imposes a predefined velocity) the induced kinematic differences are commonly assumed to be minor (Riley et al., 2007). The subjects are therefore asked to walk first without any constraint and then while wearing ski boots. The recordings are performed at two different velocities: 2.5 km/h and 3.5 km/h. The velocity is voluntarily limited as walking with ski boots on a treadmill may represent a balance challenge on its own.

B.1.2 Subjects

Seven healthy subjects from the research group participated in this study. Their main characteristics are presented in Table B.1.

B.2 Results

A general trend can be observed when comparing the BL and the locked ankle (LA) conditions data. Fig. B.1 shows the right leg articulations angles for subject S1 walking at 2.5 km/h in the two conditions (BL and LA). It can be observed that the flexion angle of the ankle is limited to about 10° in amplitude. It therefore appears that the locking is not perfect. This is due to the fact that the ski boots are not infinitely stiff (mainly for comfort reasons). A dorsi-flexion is then allowed during the weight acceptance phase. However, this flexion is limited when compared with the initial one. Moreover, the plantar flexion appears to be correctly locked. The foot remains also completely flat at all time. The ankle rotations in the other two directions are also properly locked as can be seen on Fig. B.1.

The knee trajectory is influenced by the LA condition. The main difference is an increased residual flexion angle when the heel strikes the floor. In the BL condition the knee is usually almost fully extended at this point in the cycle.

The hip appears to be less influenced by the locking than the knee. However, it can be noticed that the flexion angle is usually larger at the heel strike. Moreover, the extension phase starts a bit later in the cycle than when the ankle is free to move.

Table B.2 shows the main gait parameters for all the subjects. It can be noticed that subject 1 tends to augment his cadence in the LA condition. This observation is valid for the two different velocities. The other subjects did not exhibit any dramatic cadence changes.

The RoM of the hip is usually larger when the ankle is restrained. This is valid for all the subjects except one outlier (subject S2). Conversely, the knee movements have a more limited workspace when compared with the BL.

The changes in joints velocities and accelerations do not exhibit a clear trend. Some subjects tend to decrease one or the other while others do the exact opposite. A particular subject may even modify these parameters differently depending on the walking velocity. The differences for the maximal hip and knee velocities are –4% and –6% respectively while the difference in the RMS accelerations are +14% for the hip and –7% for the knee.

B.3 Discussion

This study investigates lower limb gait trajectories with locked ankle joint in order to specify design parameters for exoskeletons. It has been found that such trajectories are indeed different from standard normal gait trajectories. The different joints RoM, velocities and accelerations are then influenced. Nevertheless, the differences that were measured do not seem to be consistently influenced for all the subjects. Some subjects tend to decrease their maximal joint velocity and/or RMS acceleration while others augment them. These observed differences varied from –32% to +38% for the maximal velocities and from –51% to +94% for the RMS accelerations. Looking at intra-subject variations incites to conclude that the variation may be important. However, the inter-subject differences moderates this conclusion as they remain reasonable. Therefore the alterations can be considered small enough to justify neglecting them in the design specifications phase.

The gait cadence in LA augmented significantly for one subject only. The other subjects did not seem to adapt their cadence (decrease of less than 2% in average). This suggests that some individuals may adopt different strategies to adapt to the imposed condition. The small number of subjects does not allow further conclusion.

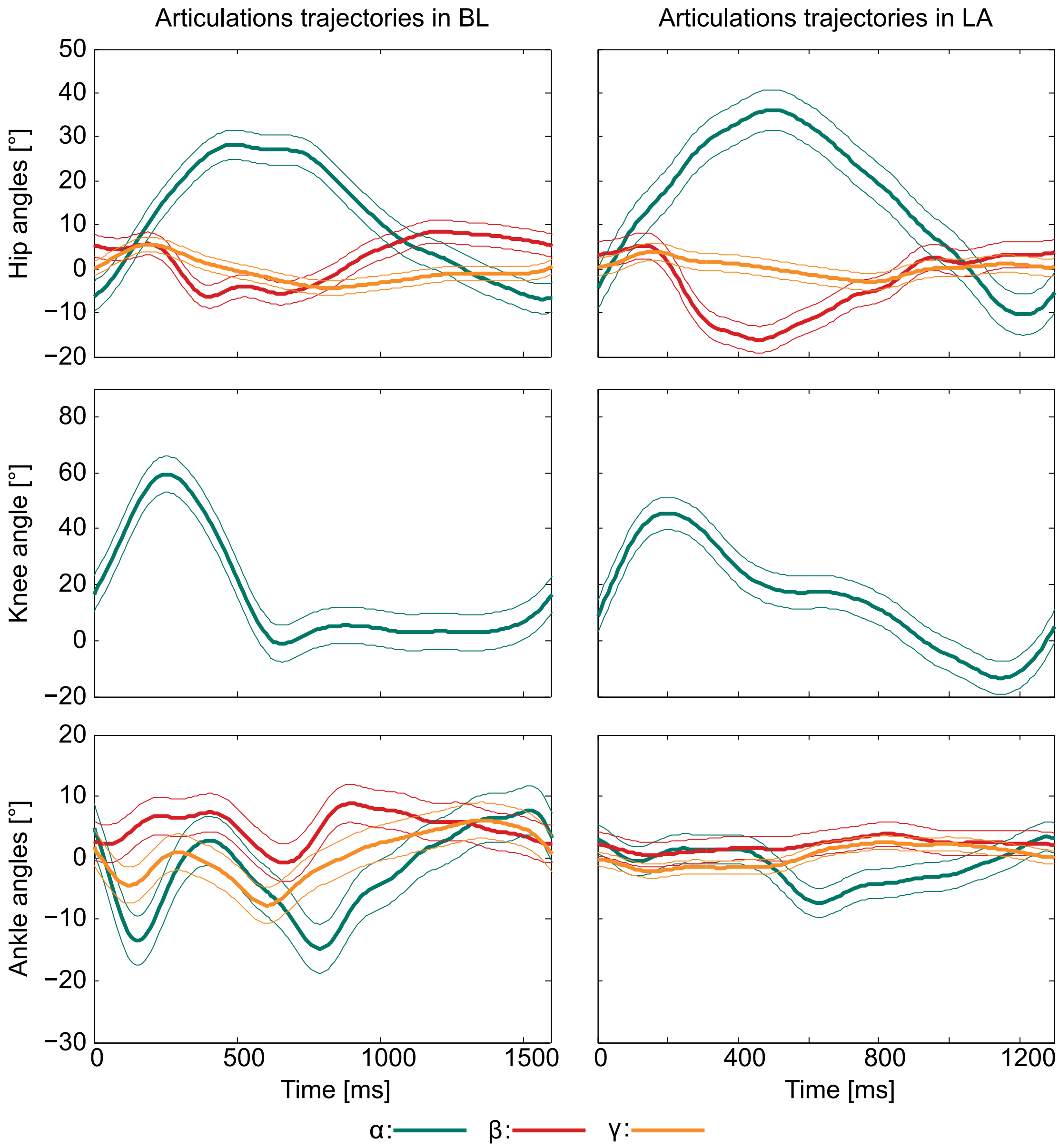


Figure B.1 – Articulation angles at the hip, knee and ankle for subject S1 walking at 2.5 km/h. Cycles start and end at right toe off. Turquoise lines are the α Euler angles which correspond respectively to the flex-ion/extension of the hip, the flexion/extension of the knee and the plantar/dorsi flexion of the ankle. Red lines present the β Euler angles and yellow lines are the γ Euler angles. The thin lines represent the 95% confidence intervals. Left: BL condition with free ankle. Right: LA condition (i.e. with ski boots).

Table B.2 – Main gait parameters for the ankle locking experiment. Two velocities were used to compare the two conditions. For each pair of parameters comparing the conditions, the largest value is in bold font.

Subject	Walking velocity (km/h)	Condition	Cadence (steps/min)	Stride length (m)	Hip range of motion (°)	Knee range of motion (°)
S1	2.5	BL	75.3	**1.106**	35.1	**60.6**
		AFR	**92.5**	0.901	**46.6**	58.8
	3.5	BL	90.7	**1.286**	40.3	**67**
		AFR	**99.3**	1.17	**45.1**	54.6
S2	2.5	BL	84.5	**0.987**	48.5	72.9
		AFR	**94**	0.886	**50.4**	**75**
	3.5	BL	**95.6**	1.22	**56**	**80**
		AFR	94	**1.24**	50.4	75
S3	2.5	BL	**75.4**	1.105	44.3	**76.7**
		AFR	74.3	**1.122**	**46.1**	47
	3.5	BL	**95.3**	1.225	49.4	**77.7**
		AFR	86.8	**1.345**	**53.5**	60.8
S4	2.5	BL	**84.5**	0.987	47.8	**67.7**
		AFR	75.4	**1.105**	**66.2**	60.3
	3.5	BL	**100.7**	1.159	52.4	**74**
		AFR	95.4	**1.223**	**63.5**	56.8
S5	2.5	BL	103.8	**0.8**	39.1	**64.1**
		AFR	**104.8**	0.795	**46**	53
	3.5	BL	108.9	**1.071**	41.2	**64.5**
		AFR	**113.8**	1.025	**54.4**	62.9
S6	2.5	BL	**90.4**	0.922	37.5	**64.3**
		AFR	86.7	**0.962**	**39.3**	45.5
	3.5	BL	**103.4**	1.128	38.9	**73**
		AFR	98.9	**1.18**	**42.4**	49.3
S7	2.5	BL	83.9	**0.993**	39.8	**67.3**
		AFR	**85.1**	0.979	**46.2**	63
	3.5	BL	**97.2**	1.2	48.5	**72.7**
		AFR	96.9	**1.204**	**50.3**	68.3

The joint trajectories are modified more consistently by the ankle and foot locking than the cadence, the velocities and the accelerations. General observations can therefore be made.
The hip flexion is larger at the heel strike in the LA condition (i.e. the extension starts later in the cycle). This enables the subjects to actually perform a heel strike and not a flat foot landing. As the angle of the ankle can only be modified within a range of few degrees (especially in plantar flexion), a smaller flexion at the hip would require a hyperextension of the knee to perform an actual heel strike. We can attribute this effect to the locking of the ankle as Browning did not report this effect at all when adding similar masses at the feet (Browning et al., 2007).
The hip RoM is also higher when wearing ski boots. This effect may also be due to the additional mass as it was also reported by Meuleman (Meuleman et al., 2013). However, the effect measured in the present study is higher than that with extra inertia only (Meuleman et al., 2013). This observation is valid for all subjects except for one outlier. A deeper analysis of this subject's data demonstrated that the locking did not work properly. Indeed his ankle flexion RoM was 20° which is largely above all the other subjects'. Consequently, this strengthens the hypothesis that the augmentation of the hip RoM is due to the ankle locking.
The knee exhibits a larger flexion at Heel Strike (HS) when the ankle is not free. This is probably due to the imposed ankle dorsi-flexion induced by the ski boots. Indeed, ski boots usually have an anterior inclination of about 10°, which may bias the results. Further investigation of this point seems required.
The results of this study have little bearing on the specifications of hip and knee exoskeleton design parameters. Indeed, the dynamics of these joints does not seem to be affected dramatically by the imposed condition (i.e. ankle restrained). Using standard trajectories from the litterature is therefore perfectly acceptable for the mechanical design phase. However, the measurement data of this study become relevant for giving guidelines concerning the control trajectories. It demonstrates clearly that the typical ankle restrictions induced by under-articulated exoskeletons have a significant influence on gait patterns. Use of healthy subjects trajectories as reference for exoskeletons may lead to ill-coordinated gait movements and thus compromise the user's safety. In order to ensure the latter, additional recordings should also be conducted over ground. The conclusion of this study remain relevant for devices that have a passive joint at the ankle as the locking technique that was used partially enables dorsi-flexion. Evidently, this should be further investigated to confirm that the trajectories are indeed comparable.

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Acronyms

3D three-dimensional. xii, 7, 10, 16, 137, 141, 142

ADL Activity of Daily Living. 2, 14, 22, 34, 46, 64, 133

AFO ankle foot orthoses. 26

AM Assistance Mode. 121–123, 125

BIC-PEA Bi-directional Clutched Parallel Elastic Actuator. 61

BL baseline. 144–146

C-PEA Clutched Parallel Elastic Actuator. 61

CAD Computer-Aided Design. xi, 71, 85, 89

CNS Central Nervous System. 14, 17, 20, 23

CoM Center of Mass. 10, 11, 114

CPG Central Pattern Generator. 14, 15

CVA cerebrovascular accident. 19

CVT Continuously Variable Transmission. 56, 61

DDO Double Differential driven Orthosis. viii, xii, 85, 88–93, 95–98, 130, 133

DM diabetes mellitus. 20

DoF Degree of Freedom. viii, xi, 5–7, 10, 26, 28–31, 47, 48, 63–67, 69, 70, 78, 79, 82, 85, 86, 102, 108, 109, 111, 112, 116, 117, 128, 140, 141, 143

EEG Electroencephalography. 132

EMG Electromyography. 17, 43, 44, 109, 125, 132

FES Functional Electrical Stimulation. 28, 30, 44

FMEA Failure Modes and Effects Analysis. 43

FSC finite-state controllers. 40

FTA Fault Tree Analysis. 43

FW Free Walking. 121–125

GRF Ground Reaction Forces. 10, 16, 21, 38, 46

HAZAN HAZard ANalysis. 43

HAZOP HAZard and OPerability. 43

HiBSO Hip Ball-Screw Orthosis. viii, xi, xii, 69, 70, 75, 78, 79, 83–85, 88–95, 97, 98, 111, 112, 114, 116–118, 121, 123, 125, 126, 130, 131

HR heart rate. 121, 123, 125, 126, 129, 131

HS Heel Strike. 8, 9, 22, 115, 119, 148

ID Identifier. 135

IMU Inertial Measurement Unit. 16, 36

IPD Idiopathic Parkinson's Disease. 20, 22

IR Infrared. 37, 135, 136

ISO International Standards Organization. 42

IVT Infinite Variable Transmission. 61, 62, 99, 102, 105, 131, 133

LA locked ankle. 144–146, 148

LED Light Emitting Diode. 135–137

LSRO, EPFL Laboratory of Robotic Systems. i, 28

MMG Mecanomyography. 17

OA Osteoarthritis. 21, 47, 127

OFS Swiss Federal Statistical Office. 1

PEA Parallel Elastic Actuators. 61

PN peripheral neuropathy. 20, 21, 46

POMA Performance Oriented Mobility Assessment. 16, 133

RCR Remote Center of Rotation. 64

RGO Reciprocal Gait Orthoses. 26

RHS Right Heel Strike. 3

RMS Root Mean Square. 49, 51, 69, 75, 95, 130, 135, 145

RoM Range of Motion. xiii, 5, 7, 16, 20–22, 26, 30, 44, 45, 47, 48, 63, 64, 69–71, 78–80, 82, 84, 98, 107–109, 112, 114, 116, 117, 123, 125, 128, 129, 145, 148

RTO Right Toe Off. 3, 137

SCI Spinal Cord Injury. 26, 27, 30, 32, 128

SEA Series Elastic Actuators. 33, 44, 57

TM Transparent Mode. 121–123, 125, 126

TO Toe Off. 3, 122, 141

TUG Timed Up and Go. 16

VSA Variable Stiffness Actuators. 57

ZMP Zero Moment Point. 143

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EDUCATION

2016	**PhD in the Manufacturing Systems and Robotics Program – Swiss Federal Institute of Technology (EPFL)** Development of walk assistive orthoses – Application to elderly people
2008	**Master of Science in Microengineering – EPFL** Specialization in Robotics
2006	Bachelor of Science in Microengineering – EPFL

PROFESSIONAL EXPERIENCE

2011 – 2016	**Teaching assistant** Robotic System Laboratory (LSRO – EPFL) Supervision of various student projects and lectures
2009 – 2011	**Hardware Engineer** VirtaMed AG (Zürich) Hardware development of diverse electronics and mechanics projects in the field of surgery simulation
2007 – 2008	**Master thesis** (EPFL) Design of a surgery tool for a medical simulator
2007	Student assistant Institut d'Optique Appliquée (IOA – EPFL) Data acquisition and image processing

STUDENT JOBS

2004 – 2007	Association des Répétitoires Ajeta (Geneva) Coach in mathematics, physics, chemistry and electrotechnology

2002 – 2006	Firmenich SA Various jobs in Supply Chain and Marketing services	

Personal Skills

Computer skills:	Software:	ProEngineer, Solidworks, Altium Designer, MATLAB
	Programming Languages:	C; C++
Languages:	French: English:	Mother tongue Fluent

Publications

Journal:

T. Vouga, **J. Olivier**, K. Z. Zhuang, M. A. Lebedev, M. A. L. Nicolelis, M. Bouri, H. Bleuler, "ΕΞΟ – A Brain-Controlled Lower Limb Exoskeleton for Rhesus Macaques", IEEE Transactions on Neural Systems and Rehabilitation Engineering, 2016, under review.

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A. Ortlieb, **J. Olivier**, M. Bouri and H. Bleuler, "Evaluation of an active optical system for lower limb motion tracking", 3D-AHM, Lausanne, Switzerland, 2014.

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K. Zhuang, T. Vouga, **J. Olivier**, M. Bouri, H. Bleuler, M. Lebedev, M. Nicolelis, "Cortical control of a lower-limb exoskeleton in rhesus monkeys", Society for Neuroscience, Washington DC, USA, 2014.

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J. Olivier, A. Ortlieb, M. Bouri, H. Bleuler, R. Clavel, "Development of an Assistive Motorized Hip Orthosis", in International Conference On Rehabilitation Robotics, Seattle, USA, 2013.

INTERESTS

Sport: Ski, sailing